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PHD

Exploration of mechanical factors in the failure of femoral hip implants

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Exploration of mechanical factors in the failure of femoral hip implants

Mark James Schmitz

A thesis submitted for the degree of Doctor of Philosophy

University of Bath

Department of Mechanical Engineering

September 2005

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Summary

The aim of this investigation was to develop suitable pre-clinical testing strategies for hip implants and elucidate risk factors that may lead to accelerated prosthetic failure.

Initially, standard pre-clinical testing methods were evaluated. It was found that testing strategies typically require prior knowledge of any worst case loading scenario. This worst case scenario may vary between designs. It was thought that the current pre-clinical testing standards were inadequate for the evaluation of novel designs, due to the inherent lack of clinical experience.

A comparative design investigation was used to explore the behaviour of the different groups of conservative implants, as these were not covered by existing pre-clinical testing standards. A broad range of conservative implants were classified by their overall geometric design. The different groups were then evaluated in a comparable manner using FEA. The results showed patterns of behaviour that were comparable to the available clinical evidence. The investigation showed that different lengths of stem may be prone to different types of failure. The more sinister type of failure scenarios seemed to be associated with the threat of loosening and bone remodelling.

Subsequent work investigated the prediction of potential failure scenarios which may occur in the future, to aid in pre-clinical testing. Bone remodelling and fibrous tissue formation were explored, and suitable simulation techniques were investigated. These methods were then evaluated using a resurfacing device and a primary stem with significant clinical experience. Comparison of simulation behaviour with clinical evidence yielded reasonable results in the majority of cases. The investigation was also useful in identifying the role of variables such as press-fit. The simulation methods were thought useful as a pre-clinical testing tool in themselves. These methods were then tested on relatively new implants, with a view to providing useful information for further pre-clinical testing strategies and radiological follow-up techniques.

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Dedication

To my wife, Louise.

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1 Preliminary investigation

1.1 Total hip replacement

1.1.1 Overview

Total hip replacement (THR) is carried out to relieve discomfort and disability caused by diseases which can disable the hip joint. The aim of the surgeon is to relieve pain, to provide motion and stability and to correct deformity, where symptoms cannot be treated with drugs. Common reasons for surgery are osteoarthritis, fracture-dislocations, rheumatoid arthritis, aseptic bone necrosis, and revision of previous hip operations. The hip joint is particularly vulnerable due to the large forces and movements involved which can cause significant pain if the joint is diseased (Martin et al. 1998). Typical forces occurring during walking are three times bodyweight, and this can rise to over eight times bodyweight during more strenuous actions such as stumbling (Bergmann et al. 2004).

Total hip replacement involves removal of the diseased bone, typically comprising the femoral head and the acetabular articulating surface, and replacing the natural joint with a prosthetic joint (Figure 1). This prosthetic joint usually consists of a ball head, attached to a stem embedded in the femur. The ball articulates against a cup in the acetabular side, thus allowing load bearing and joint movement.

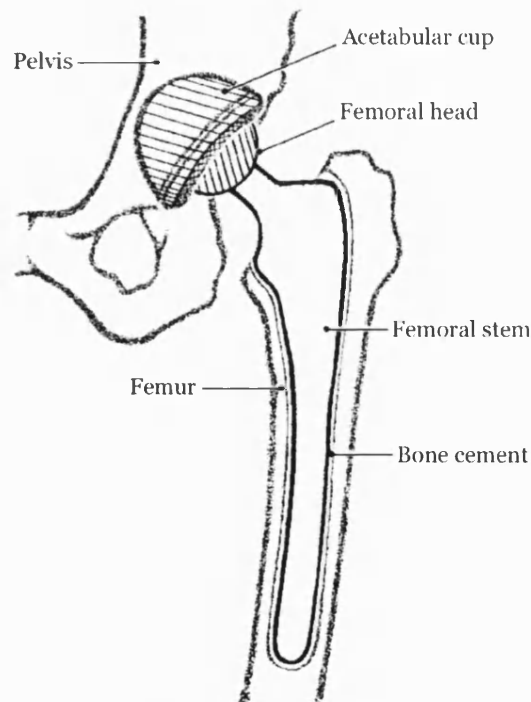


Figure 1: A typical total hip replacement [reproduced from NIH (1994)]

THR is a relatively common surgical procedure. In 2004 over 35,000 total hip replacements were performed in the National Health Service (NHS) for England and Wales (NJR 2004). Figures from 1994 (NIH 1994) estimate that more than 120,000 hip

replacements are performed annually in the United States alone - a number which has increased 64 percent in twelve years (NIH 1982), more operations of which are increasingly being done in the younger and in the oldest patient groups. THR is considered to be one of the most effective orthopaedic procedures (NICE 2000b). One of the most popular and longest established types of femoral stem used in the NHS is the Charnley (Figure 2).



Figure 2: The Charnley prosthesis (Thackray/DePuy)

1.1.2 Implant performance and revision

A number of criteria can be used to assess the clinical effectiveness of hip prostheses. The relief of pain and immobility for the patient is obviously required, and can be gauged both pre- and post-operatively. In the longer term, the main method for analysing the success of an implant is the proportion of primary THRs that require revision surgery within a specified period. To be deemed successful, an implant should have a revision rate equal to or lower than 10% after 10 years (NICE 2000b). The number of cemented femoral components requiring revision is around 5 percent at 10 years follow-up (NIH 1994). Successful implants can last in excess of 15 years, although implant lifetimes of only 5 years have been recorded in younger more active patients (Munting and Verhelpen 1993), who often have much higher revision rates (Malchau et al. 2000). Originally, fatigue failures of implants were relatively commonplace (Martens et al. 1974), (Ducheyne et al. 1975), (Wood and Timbs 1982) due to lack of control over materials production processes. Initially, stainless steel was the standard material for stems but development of surgical grade steels and then cobalt-chrome and titanium based alloys led to enhanced stem

fatigue properties. Another major cause of revision surgery is aseptic loosening (NJR 2004), but other factors such as dislocation, pain and infection are also indications for revision surgery.

A significant proportion of THR surgery is revision surgery. In 2003 around 9% of THR procedures in England and Wales were revision operations (NJR 2004). Revision surgery is required when primary implants do not function satisfactorily. Revision surgery is particularly technically demanding and is often problematic (Chandler et al. 1994) and the outcomes are known to be less satisfactory than those of primary procedures (NIH 1982). Problems include bone resorption, which can reduce support for the femoral stem and may lead to stem fracture, breakage of the stem out of the bone, or even whole bone fracture (Keaveny and Bartel 1993) with obvious consequences for the patient. Notwithstanding such problems, severe bone loss can also have a major effect on the success of the revision surgery. If bone resorption is considerable then the revision surgery can necessitate the use of bone grafting to increase the bonestock available to stabilise the implant. The restoration of lost bonestock is one of the most significant problems facing the surgeon (Heal et al. 2002). The clinical results and durability of revision hip arthroplasty with cement are known to be inferior to those of primary hip replacement (Marti et al. 1990). Revision surgery is generally much more expensive (NICE 2000a) and less successful than the primary operation and there is increased trauma to the patient. Intra-operative and early post-operative complication rates with revision surgery are relatively high, with rates of 41 percent quoted in a study of revision operations (Morrey and Kavanagh 1992). For example, if during revision the distal end of the primary implant is securely fixed in the bone, then a window in the distal bone may need to be cut to enable extraction of the failed implant (Wood and Timbs 1982). In bad cases longitudinal splitting of the femur can be required (Wilson et al. 1992), which can considerably weaken the femur.

In a survey from 1988 (Sharkness et al. 1992) 8.5 percent of implants were found to be revisions, with the predominant reason for revision being loosening, with only 18 percent of implants having been in place for more than 10 years. Rates of revision are decreasing with improved implant designs and surgical techniques (NIH 1994). At the same time many more primary THRs are being performed than before, the operation is being extended to more conditions, and to older and younger patients, and the population ages. Thus, the absolute number of revision hip replacements will increase, even if the frequency of failures in primary procedures continues to decrease. According to Dowd et al. (1995) the number of THRs is increasing at the rate of 5% annually. Dowd et al. also noted that despite improvements in materials, design and surgical technique, 15% to 20% of all THRs were revisions, of which $\frac{3}{4}$ were due to aseptic loosening.

1.1.3 Mechanism of bone remodelling

Loss of bone mass due to lack of loading is a phenomenon which has been well demonstrated by various disuse atrophy studies, including bed-rest studies and examination of changes in bone density occurring in zero-gravity situations (McCarthy et al. 2000). It is known that bone responds to the local loading environment such that if there is not enough load stimulus in a particular area then bone can resorb in that area, and equally if there is significant increase in the stimulus then an increase in bone density can occur (Cowin 1989). This phenomenon, whereby bone can adapt its architecture to the given loading situation, is sometimes called Wolff's Law, after the author of one of the first articles about the subject (Wolff 1892). The actual stimulus to which the bone responds is thought to be strain related, as this is a directly measurable quantity, although the exact nature of the process is not completely understood (Martin et al. 1998). The remodelling effect of such mechanical stimulation has also been demonstrated at the cellular level (Nagatomi et al. 1999). The effect this phenomenon can have on the long-term survival of a hip implant is quite considerable. The presence of an inherently stiff implant can mean that load is not distributed in the bone in the normal physiological manner. This is sometimes referred to as stress shielding. Concerns exist about the effect this phenomenon could have regarding progressive bone loss and consequent loosening or fracture (NIH 1994). Proximal femoral bone loss can affect the long-term survival of a replacement (Radl et al. 2000) and wasted bone stock complicates revision surgery and can affect its success. Current efforts in implant design try - as far as possible - to replicate the natural loading state found in the physiological bone, to minimise bone resorption.

1.1.4 Materials and wear problems

An important mode of long-term prosthetic failure appears to be related to generation of particulate matter. The common materials used in THR are cobalt and titanium based alloys, poly-methyl-meth-acrylate (PMMA), and ultrahigh molecular weight polyethylene (UHMWPE). The materials themselves are biocompatible in solid form, but small particles of them can cause an inflammatory reaction and subsequent bone resorption around the prosthesis, risking loosening and subsequent failure of the prosthesis in the long-term (NIH 1994). This problem can occur with both cemented and cementless components on both the femoral and acetabular side. The articulating surface between the femoral and acetabular components is recognised to be a major source of debris. Polyethylene wear particles are thought to cause an inflammatory tissue reaction. These particles have been recovered in significant quantities from periprosthetic tissues (e.g. Willert et al. 1990). Metallic debris has also been identified in significant quantities. Some of the metallic particles generated may be larger than the polyethylene debris. The major effect of these larger metallic debris may relate to the promotion of third body wear of the polyethylene, with the derivative polyethylene particles triggering the inflammatory response. However, smaller metal particles and ions have been demonstrated to be active

in direct stimulation of biologic processes (NIH 1994). Whereas wear and foreign body reaction against particulate material are important causes of long-term failure, mechanical factors are also significantly involved in implant survival. However, it is anticipated that future developments in materials technology may reduce the risk of wear debris related problems. Although it is a matter of considerable speculation as to whether any man made materials could approach the optimality and adaptability of the original undiseased biological structure.

1.1.5 Cemented and uncemented implants

Historically, the classic Charnley type cemented stem prosthesis (Figure 2) was the gold standard in THR, but after frequent problems of severe bone loss after loosening of cemented stems, increasing numbers of uncemented stems became available. Uncemented stems had the advantage that they did not use PMMA cement to achieve stability. Significant bone resorption had been associated with cemented components and attributed to cement debris. Subsequent findings confirmed that similar problems are known to occur with cementless prosthetic implants. Apart from the issue of wear debris, cement was thought to cause thermal necrosis of the bone due to the exothermic nature of the PMMA polymerisation process (Li et al. 2001). It is also known that the monomer component of PMMA is toxic, so care has to be taken over the cement preparation method. The presence of cement can also hamper subsequent revision procedures, as the old cement is difficult to remove (Pipino and Molfetta 1987). Many uncemented designs are more complex than their cemented counterparts, including features such as modular components, fenestrations, porous ingrowth surfaces and coatings, which themselves are susceptible to different failure modes (Wilson et al. 1992). Uncemented implants can also have less immediate post-operative stability than their cemented counterparts due to the lack of PMMA grouting agent. They instead rely on bone growth into porous or onto roughened surfaces for long-term fixation. Currently there appears to be no definitive answer as to whether cemented or uncemented implants perform better, each method having its own advantages and disadvantages. Studies conducted have been specific to device design and technique, and any general comparison of cemented and non-cemented stems should be viewed with caution (NIH 1994). New and improved cementing techniques have increased the success rate of THR. Techniques include the use of a stem centralising plug, a cement gun, washing of the femoral canal, cement pressurisation, and improved mixing techniques (NIH 1994).

1.1.6 Materials – traditional and potential new

Traditionally it was believed that stronger and stiffer implants were more likely to eliminate the possibility of implant fatigue failure (Semlitsch and Panic 1983). In contrast there is now increasing interest in more flexible designs (Sumner et al. 1998) to reduce the effects of bone resorption, observed with some metallic designs. Canine implanted femora

have consistently shown that cortical bone density can be significantly higher when using lower modulus prostheses (Keaveny and Bartel 1993). Typical metallic implant materials used are stainless steel, cobalt-chrome and titanium alloys which have elastic moduli of approximately 190, 220 and 110 GPa respectively. Comparing this to the typical properties of cortical and cancellous bone (15 GPa and 500 MPa respectively) it is clear that there is a mismatch (Skinner 1991). Composite materials and other low stiffness prostheses are gaining more interest in the literature (Skinner 1991, Wisniewski 1991, Evans and Gregson 1998, Soyer 1996, Christel et al. 1987, Taylor et al. 1993, Shirandami and Esat 1990, Niinmaki et al. 1994, Kaddick et al. 1996, Akay and Aslan 1995) due to the prospect of a reducing proximal resorption by increasing the load sharing to the femur. Composite structures offer the prospect of tailoring the mechanical properties of the material to suit the requirements. Non-metallic designs are also attractive because it has been noted that some patients may experience hypersensitivity as a result of release of ions from metallic prostheses, caused by wear (Liao et al. 1994). However, composites typically have lower strength than their metallic counterparts. Therefore the fatigue strength of such alternatives is obviously going to be an area of concern. Additionally, composites can have far more complex and unpredictable behaviour than traditional metallic materials. Current models of behaviour are not yet able to reliably predict mechanical performance in composites.

1.1.7 Industrial driving force for development

There are a huge range of different prosthetic devices currently on the market and under development. In England and Wales there are over 80 different brands of femoral stem in use from over 20 different companies (NJR 2004). There are many reasons for the array of different implants including; surgeon personal preference, bone morphology, bone quality, patient lifestyle, patient life expectancy and industrial interest. All these variables and more contribute to the fact that there is no single perfect implant for everyone. The orthopaedic industry is increasingly aiming to design implants which limit bone resorption and loosening and hence reduce the need for revision surgery. In the past, patients between 60 and 75 years of age were the most common candidates for THR (NIH 1994). However, the age range has been broadened to include more elderly patients, as well as younger patients, whose implants may be exposed to greater mechanical stresses over an extended period of time. A significant proportion of procedures (40 percent) are performed on patients under the age of 65 (NIH 1982), and 10-15% of all THRs are performed on patients under the age of 55 (Munting et al. 1997). A recent survey of UK surgeons indicated that 15% of the total number of THRs are being performed in younger patients (Tennet and Goddard 2000). Fatigue considerations can be expected to assume more importance in the future as younger, more active patients with higher patient expectations undergo THR. Additionally implants will need to last for longer periods due to the increasing life expectancy of patients. For example, the rate of ageing of the

American population has indicated that the number of centenarians can be expected to increase by an order of magnitude in the next 50 years (Krach and Velkoff 1999).

THR on patients over sixty years of age can provide excellent results for 15-20 years. The results for patients under the age of 50 seem to be less reliable (Munting et al. 1997) such that active patients with a life expectancy of more than 30 years will probably need revision surgery on their hips. So for younger patients, alternatives to the standard THR deserve consideration.

1.1.8 Conservative implants

Traditional stemmed implants like the Charnley have a stem which protrudes down into the central canal of the femur (medulla). This design feature is absent in the more conservative designs which apply load through the proximal region only. One of the primary aims of conservative implants is to attempt to minimise bone loss during surgery and allow subsequent easy removal and revision with a standard Charnley type implant - if required. Conservative designs are usually cementless in nature, as cement removal can be problematic during revision operations. This conservative surgery can extend the time before major revision surgery is necessary, if at all. It is particularly useful for the younger patient (NICE 2002) providing a pre-primary solution, to extend the overall life of implantability. Such designs are also becoming increasingly popular due to the ageing population and consequent increased implant survival expectations. There are many different types of conservative implants with different methods of load transfer and stabilisation, and consequently they may have different failure modes to stemmed designs. If these designs do become loose then it is hoped that bone resorption may be more limited and there may be more scope for successful revision surgery. The revision implant can be a primary device in many cases, because of the easier removal and subsequent anchorage of the new implant.

1.1.9 Testing

Pre-clinical testing is necessary for all new implants coming on to the market. Although there are in vitro tests for evaluating implant design features and material characteristics, as well as animal testing regimens, the relevance of these tests to in vivo human performance is often unknown and additional approaches are necessary. Long-term clinical studies are the only universally accepted method for evaluating the effectiveness of new designs and materials in human use. Since these take many years and are very expensive, few implant design features are supported by such thorough studies. Current pre-clinical testing standards focus on the types of failure experienced previously by standard Charnley type implants. The standard test procedures for cemented stems are presently under investigation as to their applicability to uncemented designs. Similarly it has been said that there are no satisfactory standards for the testing of composite

orthopaedic implants (Paul 1999). Currently there is no clear test protocol for the pre-clinical testing of conservative designs. If conservative designs do experience a worst case loading, it is not known what this would be, so the current cantilever stem test set-up may be unjustified and also impractical for testing practicalities. The use of appropriate pre-clinical testing methods is invaluable in identifying potential problems caused by mechanical factors and enabling them to be solved at the design stage.

1.1.10 Summary

The demands on any replaced hip are high. Individuals can take in excess of two million paces in a year, and peak loading can be over eight times bodyweight during activities such as stumbling. Therefore, any design of implant needs to be thoroughly tested. International standards exist for fatigue testing of the standard Charnley type stemmed prostheses. A conservative design philosophy for femoral implants aims to preserve as much of the original bone stock as possible to allow easier revision. The increasing number of conservative designs however do not have the conventional long stem, and instead transfer the load to the upper part of the femur. A standard test method for such conservative designs does not yet exist and would obviously be useful, both to have confidence in the resistance to failure of the device as part of pre-clinical testing and also to provide a basis for comparison between different designs.

1.2 Project aims

1.2.1 Aim

The aim of this investigation was to develop suitable pre-clinical testing strategies for hip implants and elucidate risk factors that may lead to accelerated prosthetic failure.

1.2.2 Scope

The investigation was limited to the femoral components of THR, as femoral implant geometry varies so widely that the overall design may have a significant influence on the long term success of the THR. The problem of wear was not dealt with directly in this project. Conservative implants were the main area of focus, because of renewed interest in the use of these implants, and there is little experience or guidance on pre-clinical testing strategies. A suitable pre-clinical testing strategy for novel implants was developed based on what limited clinical data was available for similar types of implants, combined with knowledge gained from in vitro investigations into the subject.

1.2.3 Objectives

There follows a list of objectives and associated reasoning for the current investigations:

To investigate the effects of implant design, using FE and clinical evidence

Implant design is an issue due to the wide variety of different designs available, and the need for an overall understanding of the effects of implant design on implant failures.

To investigate the biomechanical interactions which could result in implant failure through evolution of the biological structure

The biomechanical interactions are important but complex due to the nature of the biological environment and the wide variety of parameters which can affect behaviour.

To develop and combine numerical methods to simulate possible failure behaviour

The modelling of possible failure behaviours using numerical methods is thought to be an efficient aid to pre-clinical testing.

To validate numerical simulation methods using clinical evidence

Numerical simulation techniques need to be well validated before any significant reliance can be had in results.

To model the consequences of implantation parameters on predictive simulations

An understanding of the effects of variables, such as press-fit, on the performance of simulations may lead to a better understanding of the underlying processes.

To determine the effect of implantation parameters in the clinical situation

Clinical results can be affected by a large number of variables. An understanding of the effects of important variables on the clinical situation may be important in implant design and surgical techniques.

To gain insight into the effects of biological phenomena on the failure of femoral implants

It is important to understand how the occurrence of biological phenomena may affect the longevity of femoral implants.

To develop methods as an aid to pre-clinical testing

Pre-clinical testing strategies are required for novel implants, which by their nature have limited clinical experience to draw from.

1.3 Pre-clinical testing methods

There now follows a more detailed examination of the current legislative pre-clinical testing standards for conventional hip implants, as well as other non-standard techniques which have been used to evaluate implant performance pre-clinically.

1.3.1 Testing requirements

The standard pre-clinical test is the fatigue test. Fatigue testing is an essential part of pre-clinical testing in order to have confidence that the implant will not fail in service. This is especially relevant for hip implants because of the frequent and large loads which they have to endure over an extended period of time. If there is no generic test then it would be difficult to judge whether a new design is better/worse in fatigue than previous ones, especially where the implant designs and materials differ significantly. The fatigue test must be responsible for detecting fatigue failure of successful designs during the service lifetime. If it is known for example that stress shielding is likely to occur (as with stemmed designs) then additional appropriate testing should be carried out. Obviously this type of testing should be governed by the knowledge of probable failure modes. And as clinical experience increases the tests should be developed accordingly.

There will always be some cases of bone resorption due to the huge variations in patient activity levels, pathologies, surgical variations, and due to other factors not related directly to the mechanical design of the implant, for example wear debris related osteolysis. Implant manufacturers need failsafe confidence that when (not if) resorption occurs, the implant will not fail in fatigue, if continued implant stability is probable. An implant's capacity to obtain a stable post-loosening configuration (secondary stability) has been thought to be an important design criterion for long-term implant success (Huiskes et al. 1990). Prosthetic designs should not only be analysed with regard to their susceptibility to primary failure, but also relative to their potential to fail after the environment changes due to biomechanical changes which may occur due to the presence of the implant. Designers are continually faced with the compromise between fatigue strength and objective function in implant design, such that a worst-case test should not compromise their overall objectivity.

1.3.2 Standardised pre-clinical testing

Obviously it is vital to prevent implants from failing in service. A sensible approach to achieving this is to use some form of standardised test procedure. Such methods can help repetition of errors made on previous occasions which may have caused increased patient suffering. Such testing also allows comparability between different designs. Without proper standards widely differing testing procedures could occur between different labs, even within the same company. A number of countries have their own sets of standards but in general they tend to be very similar to the internationally agreed standards of ISO.

The current testing strategy for stemmed femoral prostheses typically involves partially embedding the prosthesis in PMMA and applying a load to the ball head. This creates a cantilever bending effect, which is thought to be a good representation of a common worst case scenario for fatigue loading of the implant. In this worst case the tip can become tightly jammed in the medullary canal, and there is no longer any significant support in the proximal region of the implant. Loss of proximal bone support could be due to several factors including breakage of the proximal bone or cement due to overloading, or resorption of bone due to underloading or wear debris. The loss of proximal support in typical femoral implants is evidenced by Sychterz et al. (2002) where the largest amount of bone loss occurred proximally in femoral stems.

The test loading is usually applied in a cyclic manner to simulate walking, because this is the most common activity, with approximately 1 million steps being taken per year on the implanted leg (Morlock et al. 2001). The waveform is usually sinusoidal in nature as this is thought to be a reasonable approximation to the gait cycle and can be achieved with most cyclic testing machines. The implant is usually tilted at an angle such that the loading creates both bending and torsional forces in the implant, as these are recognised to occur in the in vivo situation. The implant is usually expected to survive 5-10 million cycles at a load of 3-4 times bodyweight, deemed typical of in-service loading. Tests are commonly performed in Ringers solution. This is a physiological saline solution which is used to represent the potentially corrosive environment of the human body.

1.3.2.1 History of standards development

Fatigue failure of orthopaedic implants was first found to be a problem in the 1970s (Paul 1997). The increasing incidence of stem failures (reported and unreported) justified the need for understanding the factors responsible and experimental testing was started (Markolf and Amstutz 1976). Failures were commonly found to occur at a distance from the tip of about a third of the stem length (Martens et al. 1974, Ducheyne et al. 1975, Wood and Timbs 1982) caused by loss of proximal bone support and a firmly fixed distal end. Consistent observations of fatigue failure of implants were noted with failure initiated at the lateral surface of the stem (Markolf and Amstutz 1976). Initially hip joint replacements were prescribed for patients aged 65 or older, with advanced arthritic conditions. These patients usually walked with a shuffling gait, so it was thought that mechanical loading corresponding to standing on one leg was appropriate for testing (Paul 1999). Loading was initially applied parallel to the stem axis because no information was available on the loading direction at that time, but it was thought that this would provide valuable comparative information. Later studies using techniques such as gait analysis provided more information on the loading magnitudes and directions.

Much of the initial standard development was based on the work like that of Markolf and Amstutz (1976), and Semlitsch and Panic (1983, 1994) where testing of a wide range of

stemmed implants was carried out because of the need to design against fatigue failure. The testing strategy was based on the observation that there were cases of severely loosened stems that were removed after more than a year, and it was felt that the critical loadcase arose during walking, of which it was expected to see 1-2 million cycles per year on an implant. On this basis it was decided to subject partially embedded stems to sinusoidal fatigue loads in a single plane, for up to five million cycles at a test frequency of 6Hz, which corresponded to a reasonable testing time of ten days for five million load cycles. At this number of cycles the S-N curves started to show an asymptotic tendency. The work of Semlitsch and Panic was submitted in 1975 as a standard draft proposal and subsequently British and International Standards, were drawn up. The international standards ISO 7206:3 and 7 specified a form of test corresponding to this method, but these standards have now been withdrawn and superseded by more realistic tests.

1.3.2.2 ISO 7206 'Implants for surgery–Partial and total hip joint prostheses'

This is a widely adopted international standard which defines test methods for stemmed prostheses. The testing procedure has been modified and refined constantly using research and experience gained over time. The standard is divided into different parts as described in Table 1. The table shows the latest versions of the standards. At the time of writing (2005) parts 3,5,7 and 9 have been withdrawn and/or superseded.

<i>ISO 7206 Part</i>	<i>Title</i>	<i>Current Version</i>	<i>Description</i>
<i>Part 1:</i>	<i>Classification and designation of dimensions</i>	<i>1995</i>	<i>Standardises the nomenclature and dimensioning of prostheses, which is necessary to ensure comparability between different designs</i>
<i>Part 2:</i>	<i>Articulating surfaces made of metallic, ceramic and plastics materials</i>	<i>1996</i>	<i>Specifies requirements for the articulating surfaces of ball and socket hip replacements, e.g., sphericity and dimensional tolerances</i>
<i>Part 4:</i>	<i>Determination of endurance properties of stemmed femoral components with application of torsion</i>	<i>2002</i>	<i>Specifies the cantilever bending test method for determining the fatigue properties of stemmed femoral components.</i>
<i>Part 6:</i>	<i>Determination of endurance properties of head and neck region of stemmed femoral components</i>	<i>1992</i>	<i>Similar to Part 4 except that implant is embedded up to the neck resection level of the femur.</i>
<i>Part 8:</i>	<i>Endurance performance of stemmed femoral components with application of torsion</i>	<i>1995</i>	<i>Defines the fatigue loading performance required for Part 4.</i>

Table 1: The current list of ISO 7206 standards

Part 4 of ISO 7206 specifies a test method for determining the endurance properties of stemmed femoral components under specified laboratory conditions. It also defines the conditions of testing so that the important parameters that affect the components can be taken into account, and describes how the specimen is set up for testing. The basis of the current version is a form of cantilever bending test. A cyclic load is applied to the head of the test specimen (Figure 3), producing both bending and torsion, until the specimen exhibits failure or until the chosen number of cycles has been attained. The specimen is then examined for defects caused by the loading regime. The test is applicable to standard stemmed prostheses, and to prostheses designed for use in revision surgery. The standard specifies different cement depths and angles of loading for each type of prosthesis. Somewhat surprisingly, the standard does not require the testing to be carried out in Ringers solution in the latest version (2002), although previous versions have included this.

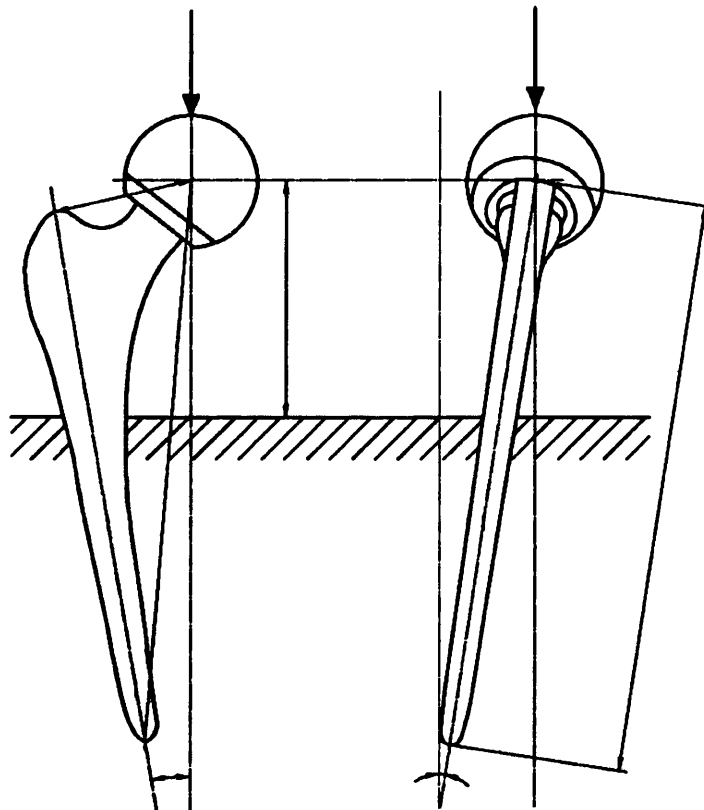


Figure 3: The ISO 7206 Part 4 test set-up (simplified from ISO 7206 Part 4:2002)

The endurance requirements, which the stems have to fulfil, are defined in Part 8. The requirement is that the implant has to survive 5 million cycles of load varying between 300 N and 2300 N. It is explicitly stated in Part 8 that the standard does not apply to prostheses for 'special clinical cases'. It is also necessary to conduct separate tests on the neck region (Part 6). This is similar to the test outlined for Part 4 except that instead of testing the worst case scenario of loss of proximal bone support, the implant is fully

embedded up to the neck resection level, as is the case without any proximal bone loss. This procedure tests the head and neck regions of the implant as opposed to the stem region in Part 4. This standard was implemented after concerns about failure in the neck regions of implants that weren't adequately covered by the cantilever bending test. The standard does not specify a load for neck testing, although work has recently been done using clinical experience and FEA to specify a suitable testing load (Howald and Bailey 2001).

1.3.2.3 British and European Standards

There are three levels of British/European standards dealing with non-active surgical implants (Table 2). These are as follows, with level 1 being the highest. The level 1 standard contains requirements that apply to all non-active surgical implants. The level 2 standards apply to a more restricted set or family of implants, such as those designed for joint replacement. The level 3 standards apply to specific types of implants within a family, such as hip joints.

<i>Standard Hierarchy</i>	<i>Identifier</i>	<i>Current Version</i>	<i>Title</i>
<i>Level 1</i>	<i>BS EN ISO 14630</i>	<i>1998</i>	<i>General requirements for non-active surgical implants</i>
<i>Level 2</i>	<i>BS EN 12010</i>	<i>1998</i>	<i>Non-active surgical implants – Joint replacement implants – Particular requirements</i>
<i>Level 3</i>	<i>BS EN 12563</i>	<i>1999</i>	<i>Non-active surgical implants – Joint replacement implants – Specific requirements for hip joint replacement implants</i>

Table 2: Hierarchy of relevant British/European implant standards

Included in the standards are sections relating to definitions, intended performance, design attributes, materials, design evaluation, manufacture, sterilisation, and packaging. It is acknowledged in the standards that the lifetime of an implant depends on the interaction of various factors. Some of these factors are the responsibility of the manufacturer, whereas others are the responsibility of the surgeon (e.g. operation technique), and some relate to the patient (e.g. the biological and physiological response to the implant, the medical condition of the patient, the conduct of the patient in respect of increasing body weight, carriage of heavy loads and adopting a high level of physical activity).

It is required that implants undergo pre-clinical evaluation by using a critical analysis of relevant scientific literature and/or analysis of data obtained from testing. It is noted that pre-clinical evaluation should consider; mechanical loads and the related movements to which the implants may be subjected when functioning, fatigue testing of highly stressed

parts, the suitability of the dimensions and shape of the implant for the intended population, and the adhesion and durability of coatings if present. The actual testing requirements refer to the relevant ISO test procedures such that femoral components of total hip joint replacement implants should be tested in accordance with ISO 7206-4 and the performance should conform to ISO 7206-8. The head and neck region of stemmed femoral components should be tested in accordance with ISO 7206-6.

It is required that implants are evaluated to demonstrate that the intended performance is achieved. Safety must be demonstrated by pre-clinical and clinical evaluation, including an appropriate risk analysis in accordance with BS EN 1441:1998 'Medical devices – Risk analysis'. This standard specifies a procedure for the manufacturer to investigate, using available information, the safety of a medical device by identifying hazards and estimating the risks associated with the device.

1.3.2.4 American Society for Testing and Materials

The ASTM F1440-92 'Standard practice for cyclic fatigue testing of metallic stemmed hip arthroplasty femoral components without torsion' (1992), is very similar to the ISO standard. The recommended test assumes a worst case situation where proximal support for the stem has been lost. It is acknowledged that the test does not address any features of a THR system that might help prevent proximal bone loss. PMMA bone cement is recommended for use as the grouting agent but any other material may be used as long as it does not chemically or mechanically interact with the specimen. The specified orientation of the implant is such that there is no torsional component induced, just bending. The test frequency is specified as 30Hz or less. No load level is specified although the ratio of minimum to maximum cyclic load, R , is specified as 0.1. There is no specified number of cycles for the test to complete, because it is noted that most materials do not possess a true fatigue limit. It is left to the investigator to determine an appropriate compromise between the amount of testing and the relationship between the test and actual device performance. A typical run-out point is quoted as 10 million cycles, with other investigations having used 5 million. Doubling the test length to 20 million cycles shows a negligible increase in data on log-log plots but doubles the testing time.

The ASTM F1440-92 provides a rationale that states that any fatigue strength data as predicted by tests following the method must be considered on a relative basis. So the tests may give valuable information about the relative merits of different prostheses, but should not be used alone as a quantitative indicator of expected in vivo performance or device lifetime. Also ASTM F1440-92 recognises that actual in vivo loading conditions should be used instead of sinusoidal loading, but due to the lack of sufficient data, standard load spectrums could not be created and hence sinusoidal loading is recommended. The testing environment is not specified in this standard, although it is said that if there is concern over the behaviour of the material in the physiological environment, a simulated

environment is not prohibited. However, if a simulated environment is used, the test frequency should be selected so as not to mask the expected effects of the environment.

1.3.2.5 FDA Guidance

The US Food and Drug Administration (FDA) has published guidelines for the testing procedures in addition to the published standards. More specifically their Guidance Document for Femoral Stem Prostheses (FDA 1995) outlines a summary of recommended test methods and reporting procedures. The rationale for fatigue testing is that it must be conducted to demonstrate that the device will continue to function without fracture in the intended patient population for an acceptable period of time. It is noted that a stem which has the same design as a previous stem except for differences in features which do not affect the stem's fatigue strength (e.g., cone taper) does not require fatigue testing provided the previous stem has passed the fatigue testing outlined. Data (e.g., stress analysis) may also be necessary to demonstrate that the differences between the old and the new stem have no effect on the stem fatigue properties. Generally it is required that testing is carried out in accordance with the methods of ISO 7206, with the following additions. Test samples should be the finished product, acceptable for clinical use. A minimum of six devices must be tested and all should survive 5 million cycles without failure. The maximum test load should equal or exceed that specified in ISO 7206. If the fatigue strength is less than that specified in 7206, then it should be equal to or better than a similar control with proven clinical performance.

Unusual stem designs or sizes may require additional testing (e.g. static three point bend test) if the clinical loading profile of the new stem differs significantly from that of standard stems. Stress analyses and mechanical bench testing may be needed to validate the test model. New stem designs (e.g. polymer composites) which fail at a load and/or number of cycles below what is described above due to new failure mechanisms, may be tested by other methods provided there is adequate clinical evidence, stress analyses and mechanical bench testing which justifies the load configurations and validates the test model; and which demonstrates that the clinical failure mechanisms of the new stem (e.g., delamination, creep, shear failure, crazing or chemical attack of polymer composite stems) would substantially deviate from failure mechanisms that would result if tested by ISO 7206.

1.3.2.6 Strengths and limitations of current standards

A problem in designing standards is that they are usually based on clinical experience; hence new designs which are significantly different cannot readily be assessed. Continuous review is carried out by the standard groups, in an attempt to keep track with current practice. This however, does not overcome the inherent problem with such historical standards. The problem is accentuated by the fact that a newly developed

implant may require extensive clinical assessment over several years before it could be determined to have any advantage over previous products (Paul 1997). Thus time would elapse before the aspects of the new design could be incorporated into the standards. In general standards reflect past clinical experience and are usually designed to replicate in vivo failures and consequently it is important to write them in such a way that they are not restrictive of new developments. Another difficulty in standard writing for orthopaedic implants is that patients lifestyles and expectations change, such that the implants which previously performed well over 10 years could now be expected to survive over 30 years at higher loads.

It is clear that there are limitations to the standard cantilever type test. The use of oversimplified boundary conditions may mask critical failure modes or induce unrealistic failures. Currently the standards specify a maximum load of approximately three times body weight, which is similar to the load encountered during walking. They do not specify an overload range to take account of events like stumbling (Bergmann et al. 2004). During the standard cantilever type tests, the part of the stem within the resin carries very little load and is not tested significantly. Also, the part of the stem above the intersection is not tested sufficiently for normal loading – hence the need for the separate neck test. The only region that is significantly tested is in the immediate vicinity of the cement interface. It is also thought that the sharp edge of the cement may in itself cause unrealistic stress concentrations. It is thought by some that current testing can be counterproductive because it encourages excessively large cross-sections and that the present methods aren't suitable for testing current or future designs, but the orthopaedic industry still agree to use them for regulatory reasons.

Time, knowledge and resource constraints dictate compromises such as; constant amplitude sinusoidal load profiles; fixed implant loading angle; test environment of ambient air or Ringers solution at 37°C rather than a complex fluid including fats, lipids, and proteins which may provide a more realistic representation of the biological environments. It was suggested by Raimondi and Pietrabissa (1999a) that the fatigue tests based on ISO guidelines led to low repeatability of results. This was explored by them using FE analysis with variations in test conditions ranging within the standard specifications. The ISO standard was thought to be a reasonable compromise between the needs of reproducibility and repeatability of a standard test and the complexity of the in vivo conditions. Raimondi and Pietrabissa (1999b) have also shown that the ISO 7206 testing specification can give experimental data of reasonable accuracy, with probably no more scatter than that found in typical specimen test results. The investigation was carried out using FE modelling, and suggested that the standard should specify a minimum thickness of embedding medium. It has been said that the loading specified in the ISO standard simulates the physiological load that is likely to occur in patients with a low or

normal body weight reasonably well, but that the test underestimates the load that heavy or active patients are likely to apply to the implant (Baleani et al. 1999).

The standard cantilever test assumes proximal bone loss combined with distal fixation, leading to a worst case cantilever loading scenario for stemmed implants. It is thought that this may be caused, at least in part, by the 'stress bypass' phenomenon (Huiskes 1993) whereby the load bypasses the proximal part of the bone in favour of the distal end of the stem. As a result the proximal femur is subject to strain adaptive resorption. This worst case phenomenon may not necessarily be applicable for new designs, that are intended to increase loading on proximal femur and/or do not have a distal stem. For stemmed designs which transfer load proximally, there is an argument that the distal part of the stem should not be so highly loaded. Obviously, if such a proximal fixation method fails, then the test justification for the normal stemmed devices would apply. For new types of composite implant designs, load sharing with the femur is a design goal and complete loss of bone support is much less likely, so there is an argument that the cantilever test becomes obsolete in these cases. The standard stress bypass test is also not applicable for conservative designs which may have entirely different types of failure scenarios to stemmed devices. Another drawback is that the nature of the test load is set by clinical experience of failures in previous implants, although this may not necessarily be applicable for new designs.

It is thought that the current standards are inadequate for the evaluation of composite materials, modular designs and systems incorporating proximal support (Postak et al. 1991), and will not adequately describe in vivo loading conditions with future implants (Styles et al. 1998). Therefore, there is a requirement for test method development. Current testing standards only test one or two modes of failure, and other modes may become apparent in the future.

The main issue with fatigue testing is that to be of benefit for new implants it requires prior knowledge of any worst case loading scenario (such as cantilever bending) which may be likely to arise due to adaptation of the biomechanical environment. This can partly be inferred from experience with similar implants, but cannot ultimately be judged until clinical trial results for a given implant are available.

1.3.3 Further testing methods

1.3.3.1 *Measurement techniques*

Knowledge of the effect the loaded implant can have on the surrounding bone is implicit in understanding the behaviour of the construct. Strain measurement is an effective method for evaluating these stress distributions in the femoral bone. Three principal techniques are available which are strain gauge measurement, photoelastic methods, and finite element analysis.

Strain gauge measurement is by far the most direct measurement method, with magnitude as well as direction being calculable for a single plane using rosette strain gauges. The measurements are limited to discrete points and are only realistically applicable to the surface of the femur, although implant surface strains can be measured when embedded within cement. This experimental technique is commonly applied in vitro test set-ups, where the implant is typically implanted in a cadaveric bone, which can then be loaded in a rig which approximates the loading conditions in the biomechanical environment.

The photoelastic method is a novel experimental technique. It utilises a photoelastic coating which responds to variations in stress by showing different coloured fringes. This in vitro technique is useful in that it can provide a continuous (rather than discrete) description of the strain behaviour of the surface. This technique is limited in that it is only semi-quantitative, although investigations are ongoing into computational analysis of the results (Morris et al. 1999). The photoelastic method can also only indicate quantitative strain behaviour in the plane of the surface, although polarised light analysis of loaded 3D translucent plastic models can give an idea of the overall strain behaviour throughout a body (Lau and Teoh 1999).

The finite element method is a computer based approach which analyses a numerical approximation to the system in question. It requires descriptions of boundary conditions loads and materials properties to derive the behaviour of the system. A distinct advantage of the FE method is that it allows visualisation of the magnitudes and directions of stresses and strains throughout the whole structure, and parameters and variables can be strictly controlled. However, the accuracy of the output is highly dependent on the accuracy of the information supplied, and some form of model validation is usually required. It has been seen that FE based pre-clinical testing can give meaningful results (Stolk et al. 2002a, 2002b).

1.3.3.2 Non-standardised test set-ups

There now follows a description of various testing set-ups which have been explored and used in pre-clinical testing of femoral hip implants.

Cadaveric femora

Testing prostheses using cadaveric human femora may appear the most realistic testing scenario, since the material properties and geometry would be identical to those found in vivo. But use of cadaveric femora presents significant problems in terms of their availability and their variability. Not only can the geometry of individual bones vary considerably, but also the mechanical properties of the bone. The properties are not only affected by the state of health of the patient's bone prior to death, but also the way in which it has been stored (fresh, frozen, embalmed, dried) prior to testing (Dobbs and Chaplin 1981). It is also known that when in vivo, small fatigue fractures may be repaired

in the normal course of bone remodelling, whereas in vitro the fatigue performance is somewhat reduced. Thus studies performed using cadaveric bones are usually limited to static loading. Christel et al. (1987) used strain measurements from a statically loaded implanted cadaveric human femur and compared the results with published material fatigue strength data to get an idea of the safety factor for the implant. Munting and Verhelpen (1993) developed a detailed loading set-up for a cadaveric implanted bone to monitor implant stability. Human femurs have often been used for in vitro tests of THR components, but the variability of cadaveric specimens has always been a significant problem. It has been estimated that the inter-femur strain variability is sometimes in excess of 100% of the mean (Cristofolini et al. 1996), requiring enormous sample sizes to obtain satisfactorily significant results.

Synthetic femora

Composite bones have been investigated as in vitro test subject because of the problems associated with cadaveric material. In terms of availability there is only one brand of commercially available composite replicate femur known at the time of writing. Pacific Research Labs (www.sawbones.com) have produced three generations of composite femur or 'sawbones'. Sawbones are constructed with glass fibre reinforced epoxy resin to represent cortical bone, and cancellous bone is represented by rigid polyurethane foam. They have approximately the same morphology as a femur and aim to replicate the overall mechanical properties of a femur. The first generation was constructed from a carbon-fibre reinforced epoxy shell with polyurethane foam core. The second generation used glass fibre reinforced epoxy shell formed by cloth, matt, and roving with a polyurethane foam core. The latest, third generation, composite replicate femur uses injection moulded short glass fibre filled epoxy around a polyurethane foam core. Composite femurs are often used in comparative testing of hip implants for static and micromotion studies, as the bones are designed to replicate the material modulus and geometry of human cadaveric bone, and have much lower inter-femur variability and are easier to handle, with ready availability. Their reasonably realistic properties, availability and relatively low cost are appealing for use in comparative primary stability studies.

It has been shown that sawbones mimic the design and properties of the human femur (Szivek et al. 1990). It has also been shown that the first generation composite femurs had highly variable torsional and bending responses (Szivek et al. 1990). Szivek and Gealer (1991) investigated the behaviour of the second generation composites and noted that their stiffness was generally low in comparison to that measured from cadaveric bones. An investigation into the properties of the third generation of composite femur (Heiner and Brown 2001) showed that the new design was considerably less stiff than natural human femurs in the more proximal regions. There appears to have been little or no dedicated investigation into the fatigue behaviour of composite femora.

However, drawbacks of their use for fatigue testing have been highlighted because of significant time investment in specimen preparation, and tests have shown that the composite femurs are unable to withstand long-term elevated load (over 2000N) with subsidence and splitting of the cortical wall being evident during fatigue testing (Humphrey and Gilbertson 1993). It appears that composite synthetic models are suitable substitutes for cadaveric specimens for the purpose of comparative primary stability investigations, but not suitable as a standard fatigue testing subject. Further discussion on cadaveric/synthetic femora can be found in the review by Cristofolini (1997).

Supporting/embedding methods

Krygier et al. (1994) and Postak et al. (1991) used similar fixtures for testing proximally modular stems, whereby proximal fixation was simulated using a potted proximal area, along with a simple distal support, both tests being performed in air at room temperature. Postak et al. (1991) developed a test set-up whereby the proximal part of the stem was fixed in a polymer and the distal part of the stem was simply supported, although exact test details are limited. The jig developed by Krygier et al. (1994) used PMMA as a proximal grouting agent. It was noted that the results of these tests could not be compared with fatigue data on non-modular stems tested in more conventional manners. The method was tested on various different designs of proximally fixed stem by Heim et al. (1995) using a decreasing load until 10 million cycles was achieved, indicating the endurance limit of the implants. Acrylic was used as the embedding medium. Results showed that fretting played a major part in the failure process. The set-up enabled simple test duplication and is amenable to environmental testing, but only provided single plane of transmission of loads.

Maharaj and Jamison (1993) carried out creep testing on a composite hip stem using a titanium support fixture with a contoured gap starting tangentially in the medial midstem and extending proximally to a maximum separation of 3mm. A similar supporting test set-up fatigue testing composite stems was detailed by Wisenewski (1991). A three inch diameter titanium cylinder was used with a urethane insert or with titanium displacement control inserts. Evaluation of the fixture with strain gauged implants showed that the urethane insert produced similar implant strain measurements to those measured in an embalmed human femur, although use of titanium inserts did not give as good results. These designs had the disadvantage that they did not produce torsion in the implant. Another disadvantage of a such test set-up is that a new jig would need to be created for each different implant design, and comparability between implants would therefore be difficult. Complete encapsulation of the prosthesis complicates monitoring of the test specimen for debonding/degradation/fracture and also restricts contact with the testing environment. These tubular designs were easy to fabricate and the uncomplicated geometry resulted in simple test set-up and repeatable test conditions. It was relatively

low in cost, apart from the metal insert manufacture, and could accommodate changes in orientation of the implant relatively easily.

Bending tests

Ducheyne et al. (1983) outlined a method of testing differing from the normal load-on-head configuration whereby the stem was loaded in four point bending. He ran tests in air at 30 Hz for 10 million cycles. The main advantages of such a method was that there was good reproducibility and the stems critical section was not predetermined by cement depth, and hence the method was proposed as a standard test. This method can be carried out relatively fast and it is not hampered at such speeds by the behaviour of polymer potting mediums at high frequency, although the loading may not have been representative of the physiological situation.

Akay and Aslan (1995) used a combination of methods to get an estimation of the fatigue life of a CF/PEEK prosthesis. Mechanical properties including modulus, strength and fatigue limit were calculated from 4 point bend tests of bar specimens in 37°C saline. Fracture toughness and fatigue crack growth rates were measured using notched bar specimens in 3 point bending. FEA was also carried out to determine the worst stresses in the implant. This was combined with the fatigue limit determined from the 4 point bend tests to determine a factor of safety for the implant, which showed fatigue failure was unlikely. A fracture mechanics approach was then taken using the other data collected combined with information about the average size of voids in the prosthesis, which calculated that the growth of the voids was unlikely.

Finite Element Analysis

Finite element methods can be used to predict the mechanical performance of different devices on a comparative basis. An overall view of the stresses being developed in the structure is typically obtained, so the method is a relatively powerful tool. Stress analyses can provide guidelines to assess consequences of design choices and surgical compromises and can add to the understanding of the complicated bone/implant structure by quantifying and correlating variables and observations. For example the effect of parameters such as cement layer thickness, stem stiffness, stem length, and stem cross-sectional shape have been established and quantified (NIH 1982).

Finite Element Analysis (FEA) can be a useful tool, and its use in endurance verification of implants has been investigated, especially with regards to custom made hip prostheses (Baleani et al. 2000, Viceconti et al. 1998) where variable implant features and fast turn-around are important. A 3D FEM can be developed which predicts the stresses and strains in the loaded THR, although model realism can be limited by computing power, interpretation of boundary conditions, and materials properties. The latter is important

not only due to the highly complex biological structure that is bone, but also because of the lack of understanding of behaviour of new materials such as composites.

FEA has been used extensively to investigate the effects of THR. Raimondi and Pietrabissa (1999b) used FEA to investigate the effects of varying different parameters associated with the ISO 7206 standard. Baleani et al. (2000) also created a representation of the ISO 7206 (1989) standard. Taylor and Tanner (1995) showed that varying the coefficient of friction between bone and implant in FE modelling had significant effects on results. Yildiz et al. (1998a, 1998b) developed an FE analysis to examine the behaviour of a laminate composite prosthesis in a femur and it was shown that more favourable stresses and deformations could be generated in the femur using composite implants compared to conventional metal ones. It has been shown that FEA can effectively be used for the design evaluation of hip prostheses prior to prototype production, and that trends in lab experiments can be reproduced using FE (Verdonschot et al. 1993). FEA can also be used to produce information about the stress shielding of the bone, interface motions and stresses at modular prosthetic connections. It is thought that pre-clinical testing using the FE method is best applied on a relative basis in standardised bone models (Huiskes 1993), and it is thought that FEA should be complemented by laboratory bench testing to remove some of the inherent uncertainties of computer models, once a prototype of the prosthesis is available. FE models can also be used in the construction of physical testing regimes, such as aiding in the placement of strain gauges. Even though the realism of computational models is becoming increasingly credible, it is recommended that results should be verified by testing (Baleani et al. 2000). FEA is often carried out on new designs to analyse behaviour by establishing the effects of innovative design features and to test pre-clinically the safety of devices relative to failure scenarios that are well defined (Huiskes and van Rietbergen 1995). These studies are often done in isolation for each design and so are often of limited general use.

1.3.3.3 Comparison of further testing methods

Several of the testing methods have significantly simplified boundary conditions including the bending and supporting methods, which do not seem very physiologically similar. The cadaveric and composite bone systems have a more realistic load sharing ability. Of the various testing procedures described above FEA seems to offer the most potential for evaluation of the impact of an implant on the biomechanical environment, as design variables can easily be changed, and the effects can be monitored throughout the whole bone. The accuracy of the simulation is controlled by the level of refinement of the materials properties and boundary conditions. FEA does have the requirement of validation, but this can be addressed with comparison with available clinical data. Fatigue issues may be identified using FEA, but definitive physical testing should be carried out if possible and co-ordinated using the information gained during FE analyses.

1.3.4 Summary of pre-clinical testing methods

It is clear that there is a requirement for pre-clinical testing to detect possible failure of designs during the service lifetime. Fatigue testing forms the mainstay of current pre-clinical testing. Tests to simulate the biomechanical environment have been developed in a variety of forms which vary in their complexity. These tests typically simulate simplified loading and boundary conditions, approximating to the environment encountered in vivo. If it is known that non-standard loading conditions are likely to arise, then additional appropriate testing should be carried out for these cases. The main issue with fatigue testing is that, to be of benefit for new implants, it requires prior knowledge of any worst case loading scenario (such as cantilever bending) which maybe likely to arise due to adaptation of the biomechanical structure. This worst case scenario may vary between designs. It is thought that the current pre-clinical testing standards are inadequate for the evaluation of novel designs, and new materials, due to the lack of long term clinical experience. Another major flaw of the pre-clinical fatigue test is that it does not adequately address some of the major causes of implant failure, such as those associated with implant loosening and bone remodelling.

2 Implant design investigation

2.1 Aim

The aim of this investigation was to explore the behaviour of the different groups of conservative implants. The study evaluated a broad range of generic designs in a comparable manner using FEA. This was then combined with the relevant clinical information available, to give a greater understanding of the behaviour of such systems. The results were then analysed to establish any potential risks relevant to any particular type of device.

2.1.1 Investigation outline

Previous comparative investigations between different implant designs were first investigated, to understand the benefits of the various techniques. It was recognised that implant designs can vary widely, and can contain many features. The functions of the various design features were hence investigated and discussed. The implants were then classified into various subgroups to aid in further analysis. For each subgroup the relevant clinical behaviour was investigated. FEA was then carried out on the various implant groups to try and establish any patterns in behaviour, and establish comparisons between designs. This behaviour was then compared to the clinical data previously collected, with inferences being drawn between the implants mechanical and clinical behaviour.

2.2 Previous comparative biomechanical evaluations of implants

There now follows a review of other studies which have compared the performance of a series of devices.

A theoretical biomechanical evaluation of several types of femoral implants including stemmed and conservative designs was carried out by Haboush (1953). All the implants investigated were considered to be successful under ideal conditions. The analysis was conducted on the premise that the implant would become unsupported in some areas due to various factors including bone remodelling, operative variations, implant variations and patient variations, with the worst possible case depicted for each implant. Since exact data pertaining to each individual implant could not be determined, it was said that reasonable assumptions were made in each case. Given that the analysis was theoretical in nature with associated assumptions, it was noted that the primary observations would be relative in nature, since the same assumptions were applied to all. Contact pressure was used as a descriptor of the performance of the device – where a uniform pressure distribution was desired, and concentrations of pressure were to be avoided. It was acknowledged that hoop stress in the femur was ignored in the analysis but could be critical due to bone's low tensile strength. The method highlighted various potential problems with different devices, although the analysis seemed somewhat biased in favour of the author's design of implant.

Gruen et al. (1979) carried out a retrospective evaluation of 389 traditional stem type THRs to identify potential modes of failure, as an aid to classifying the loosening of cemented femoral components. The femur was delineated into 7 zones around the cemented femoral component (Gruen zones) and these zones were examined for radiographic evidence of loosening and classified as one of 4 modes of failure (pistoning, medial mid-stem pivot, calcar pivot and bending cantilever). These modes of failure indicated a deviation from the stable femoral component where there was adequate fixation and load bearing for each interface for the Charnley type stem. The frequency with which each type of mode occurred was assessed. The consequences of each mode of failure were evaluated such that the cantilever bending case was seen to be the most sinister with respect to stem durability. Some modes were progressively symptomatic whilst others were seen to be arrested with no further symptomatic problems. It was noted that other types of implant may have different types of failure modes.

Markolf and Amstutz (1976) carried out an experimental study to investigate stem failures. Implant surface strains were measured during various loading conditions at twelve locations in three commercially available hip stems. The stress magnitudes and distributions in the components were compared in each component and the effect of varus/valgus load inclination on these results was investigated. The effects of loose and also partial fixation were also investigated. The implant with the highest stresses corresponded to the implant with the most failures and in the same region as that found during testing. This study was useful because it allowed comparison between different designs and highlighted potential weaknesses in implants which could occur clinically and helped to explain why prostheses were failing through loosening, lack of support, and materials problems. Partially fixed prostheses, similar to the cantilever bending used in current standards, showed the highest stresses of all the configurations tested, highlighting the danger caused by loss of proximo-medial support by cement fracture or bone resorption. Varus placement of the stem was also recommended to be avoided because this increased the stresses on the stems, as did moving the stem before the acrylic had set. It also highlighted shortcomings of some materials used in prosthesis manufacture.

2.3 Implant features and their functions

There now follows a discussion of various notable implant features which can be found on many implants. Their method of function and associated risks are discussed.

2.3.1 Collars

Collars can offer direct compressive geometric load transfer to the resected neck of the femur. Most of this loading is transmitted through the stiff femoral cortical bone in a similar manner to the loading through the physiological femoral neck, although the intact femoral neck would also transmit some tension in bending in the lateral region. Collars transmit the majority of the load in a single plane, perpendicular to the resected femoral neck. In traditional stemmed implants it has been noted that collars are important for load transfer (Whiteside and Easley 1989). Bone can respond well to this direct loading with corresponding increases in bone density underneath the collar (Huggler and Jacob 1995, Munting et al. 1997). The collar can also prevent axial subsidence of the implant and axial micromotion (Whiteside and Easley 1989) by virtue of the direct geometric impedance of the implant by the stiff cortical bone.

In an investigation of stemmed implants, a large change in loading patterns occurred with the loss of collar support (Keaveny and Bartel 1993). It was thought that if the collar was unable to transmit enough load to the adjacent bone, resorption was likely. Possible reasons for inadequate loading of adjacent bone include inadequate initial contact with the implant (Munting et al. 1997) due to incorrect positioning or bone preparation. If inadequate collar loading to the resected neck does occur, the situation is unlikely to return to normal, as the bone will tend to favour bone growth towards the current preferential load path, although stabilisation may occur through sufficient bone growth towards the non-design load path. This is illustrated by two different, but stable, remodelling patterns occurring with the Thrust Plate Prothesis (Taylor et al. 2004). A possible reason for inadequate loading of the adjacent bone would be due to unforeseen load transfer to other regions. This has been observed particularly in traditional stemmed implants whereby rigid fixation of the distal stem and high implant stiffness relative to the bone, has created a preferential load path away from the proximal region. The rigid distal fixation may be created during surgery (intentionally or otherwise) or may result from a subsequent remodelling response of the host bone. Indeed any area of implant fixation which tends to attract the desired loading away from the collar region renders the construct prone to the effects of subsequent remodelling. Insufficient stability is another possible reason for the failure of the construct, as excessive micromotion can cause the interface to resorb leading to a similar scenario to inadequate initial contact, although restabilisation of the implant is less likely due to the current and inevitably decreasing amount of stability.

2.3.2 Tapers

Tapers are used as a method of load transfer. The wedging effect transfers load partly through shear force and partly through compressive force normal to the taper surface. The relative amount of these forces depend on the frictional interface conditions between the taper and surrounding bone/cement (Howie et al. 1998) and the angle of taper, and the angle of loading (Kuiper and Huiskes 1996). In the surrounding bone the taper creates compressive stress in a radial direction and tensile hoop stress in the circumferential direction. A double taper is typically employed coincident with the medullary canal axis or the main axial load direction. Tapers can also transmit loads directly to bone through the variety of bending loads which occur during patient activity. The taper has the advantage that it is a self correcting system such that many problems which may occur with the seating and load transfer tend to correct themselves. If there is inadequate or eccentric fit between implant and bone the taper can subside leading to a re-centring and retightening of the implant in the construct. Thus, some degree of subsidence is commonly associated and desired with tapered devices. Tapers are used with both cemented and uncemented constructions.

Significant stem subsidence is not ideal, quite apart from the effect it can have on leg length. It has been noted that high stem migration is a predictor of early failure of cemented stems (Verdonschot et al. 2002). In study of a tapered stem design, van Rietbergen et al. (1993) concluded that subsidence could eventually lead to distal jamming and stress bypass. This was supported by the experience of a small tapered stem that tended to subside leading to more cases of proximal osteolysis than distal osteolysis (Morrey et al. 2000). Conversely, any feature that tends to prevent significant subsidence of the taper could reduce the loading of the bone surrounding the taper, which could lead to resorption of bone around the taper itself. Another possible failure mechanism of the taper is if the direct stress, shear stresses or micromotion are continually too high at the interface of a taper then bone can resorb and continued subsidence can occur which can cause the patient problems due to leg length or pain requiring revision.

2.3.3 Stems

The stem is primarily a feature for geometric interlocking within the medulla. In traditional stemmed implants it has been noted that collars are important for load transfer (Whiteside and Easley 1989) whereas the stem is important for stability and the elimination of toggle. The primary loading aim of traditional stems is generally direct radial loading to the inside of the cortical bone of the medullary canal. This gives a long moment arm to counteract the bending loads which occur through the implant. With shorter stems the load is either transmitted to the cancellous bone in the metaphysis or the lateral cortical bone acting as the other end of the 'see-saw', where the pivot point is

usually the proximo-medial cortical bone. Stems can also help transmit torsion, because of the offset of the tip from the main load line.

The way in which the stem transmits load is important. If an intramedullary stem transmits significant axial load (either due to under-reaming of the canal or inappropriate surface coatings etc.) then the stress bypass phenomenon is likely to occur, detracting load away from the main load-path. This can lead to increasing density in the distal region and loss of bone in the proximal region, a situation which can cause undue stresses the implant. Proximal resorption of bone due to stress shielding has been observed to occur significantly more often with distal press fitting stems (Engh and Bobyn 1988). Stem curvature can have an effect on loading behaviour (Berzins et al. 1993) as can stem length (Tanner et al. 1995). It has been said that reducing the stem length may have an adverse effect on bone-implant interface stability (Joshi et al. 2000).

2.3.4 Bolts / tension devices

Tension devices are not usually the main load transfer mechanisms as the load transfer required to the bone is primarily compressive. Tension devices are used to anchor the major load transfer component in place. They carry mainly tension loads and provide opposing stability, usually affixed to the stiff lateral cortical bone. The devices can also help transmit bending loads across the trochanteric region.

Potential problems with such devices are that significant loads can be transmitted through them, so they can fail. This was demonstrated by fatigue fracture of the main tensioning bolt several times in developmental testing phase of a new implant of this type (Munting and Verhelpen 1995). The primary tensioning applied in such devices during surgery may subside (Bereiter et al. 1991), but this does not seem to be detrimental to long term stability.

2.3.5 Fins / pedicles

Fins and similar small geometric structures are often implemented to increase resistance to torsion. Torsion is created due to the offset of the head of the implant and the angle of loading, particularly during activities such as stair climbing. The features function as a torsional load transfer mechanism and they also provide increased stability. Features are usually impacted into cancellous bone to provide an initial frictional fit. Potential problems with such features are they are relatively small and may act as stress raisers in the implant. Thus, care has to be taken to limit their effect on the fatigue life of the prosthesis.

2.3.6 Shape / fit and fill

Increasing the congruency between implant and bone is a method that has been used to provide increased stability to an implant. Adjusting the areas of fit can distribute the

applied load in a desired manner (Kim et al. 2001), e.g. a better proximal fit can increase the loading to the proximal femur. The aim behind increased fit and fill is to distribute the load over as wide an interface as possible so that no part of the bone resorbs due to stress shielding. If a good fit can be achieved then micromotion can be minimised. Internal fitting devices are commonly inserted into a specially reamed cavity in the bone. External fitting devices, such as resurfacing prostheses are fitted onto a specially prepared surface such as the resected neck of the femur.

Cement is commonly used as a grouting agent, which can create an excellent match to the contour of the implant whilst allowing interdigitation with the surrounding bone. The bone-cement interface is usually good, transmitting direct (compressive and tensile) and shear forces at all points of contact. However, loadbearing can be poor if there is limited interdigitation. The cement-implant interface allows transmission of direct compressive and a small degree of shear only. The grouting agent also acts as a mediating layer between the stiff implant and relatively soft bone, distributing stresses more evenly. Uncemented implants usually try to achieve a similar fit, but it is not as easy to achieve because of the lack of a grouting agent. Custom implants are commonly designed for special cases to achieve the best fit possible where standard THR is not possible. Cancellous bone is usually removed and the implant is designed to contact the endosteal surface. Interference fits are usually sought, as this provides a more secure contact and increases primary stability. An in depth analysis of current literature regarding fit and fill is documented by Laine (2001).

2.3.7 Surface structure

It is recognised that the surface finish (grit blasted or polished etc.) affects the frictional behaviour between the implant and bone, and can have a marked effect on load distribution (Keaveny and Bartel 1993). The position of surface structures is also critical as variations in position can change the load path and lead to bone growth/resorption (Engh and Bobyn 1988). Many geometric surface treatments (e.g. sintered bead coatings) aim to achieve a 'macro-lock' with bone interdigitation into the surface structure. Such constructs can transmit significant shear and direct loads (compressive and tensile) up to the failure loads of the interdigital bone. These geometric macrolocks take time to establish because of the time taken for the bone to grow into the structure, so they are not usually the initial primary method of load transfer. Some surface treatments are non-geometric in nature. The microroughness of surfaces (roughness below 50µm) can affect the fixation of bone and connective tissue due to the change in adhesion which affects atoms ions biomolecules and cells. This subject is dealt with in more detail by Wen et al. (1996). Biochemically active treatments such as hydroxyapatite (HA) can be used to enhance the attractiveness of a structure for bone apposition. Although the geometric method of load transfer will ultimately determine the long-term success of the integration of the implant,

the importance of a good start cannot be underestimated (Rasquinha et al. 2002). It is thought that HA coating cannot substitute for stable mechanical fixation (Lai et al. 2002).

A major problem with relying on surface interactions for fixation is that they take time to establish because the bone has to grow onto or into the structure and a certain level of stability is required before the interaction can occur. Micromotion needs to be below a threshold of about 20 μm before interdigitation can occur (Büchler et al. 2002). The use of surface treatments is also not without potential risk, as there is concern over the durability of surface coatings (Rokkum et al. 2002) and also the stress raising nature of irregular surface structures and surface treatment processes. It is also known that HA is progressively resorbed (Radl et al. 2000) and failure of the HA coating itself or either of the interfaces can also occur.

2.3.8 Discussion of features

It can be seen that different design features have different functions which can often be related to either bone mechanics or interface mechanics. Bone mechanics requires the transfer of load from the prosthesis to the bone in desired areas and directions in order to maintain normal bone mass. Interface mechanics require the stabilisation of the implant in terms of micromotion or subsidence – to promote implant integration by direct bone ongrowth or ingrowth. It has been said that the design principles for optimal interface mechanics (low micromotion - rigid, fully bonded, long stem) can conflict with those requirements for optimal bone mechanics (good proximal load transfer - flexible, short stem, proximally bonded) (Huiskes and van Rietbergen 1995). With a stemmed prosthesis, the moment which arises from the load at the prosthesis head, is typically counteracted by the reaction force acting in the neck region in combination with the normal force acting at the stem tip. If the stem is shortened, the distal reaction force is moved proximally and increases owing to the shorter moment arm. Theoretically, reducing the stem length could reduce stress shielding, but at the cost of reduced interface area for load transfer leading to higher stresses and interface failure probability (Huiskes 1993). The interaction of these features in their respective functions is also likely to be important. Features such as that the collar and the taper may seem mutually exclusive in terms of a primary geometric method of load transfer, as the use of a collar would not allow the taper to perform properly, and vice-versa. This may be true in principle, but it is noted the Lubinus SPII has both a taper and a collar and yet achieves very good results (Malchau et al. 2000). It is apparent that the performance of any of these features is dependent on many other factors including the skill of the surgeon and the response of the host bone.

2.4 Classification of conservative devices

There are many different types of conservative devices on the market. In this investigation they have been divided into groups to aid further analysis of the implants in general. Since features such as collars, tapers, fins etc. are possible features on all types of implant the classification was based on overall shape and structures used to achieve primary geometric stabilisation.

2.4.1 Type 1: Resurfacing

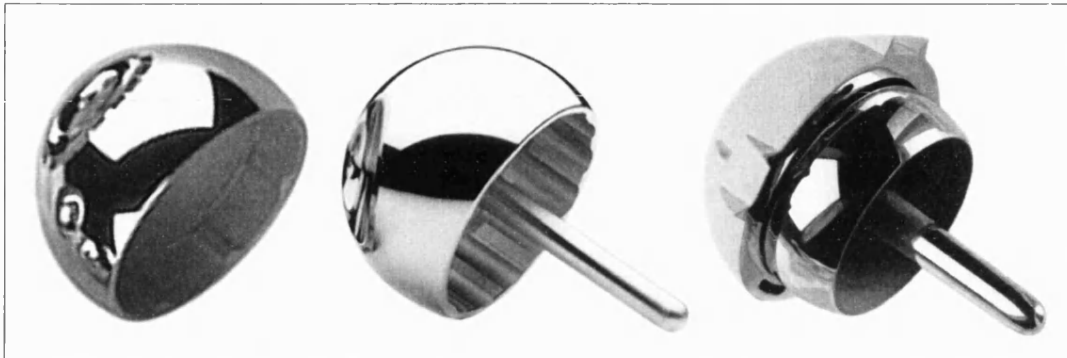


Figure 4: Images of resurfacing type implants (from left: Wagner¹, Durom² and Cormet³)

Description

A resurfacing prosthesis replaces the surface of the femoral head. Implants are typically fitted externally onto the prepared surface of the resected neck of the femur. They may have a central guide pin for aiding placement during surgery, although this is not typically a primary load bearing feature.

Structural functionality

The implant is loaded directly from the acetabulum and the load is transmitted to the underlying bone. The implant is mainly under compression, although shear and separation forces may occur at the interface due to the relative stiffness of bone and implant. This produces bending in the femoral neck leading to compressive forces in the medial and tensile forces in the lateral femoral neck and should create a more physiological load transfer in the major part of the femur below the implant. The distance from load application to anchorage point is minimal so stabilisation is dependent on the limited surface area available and is achieved using the geometric interlock between the femoral head and implant typically using a layer of cement as a grouting agent to resist shearing action, or a contoured internal surface.

¹ Zimmer GmbH

² Zimmer GmbH

³ Corin Group

2.4.2 Type 2: Transtrochanteric

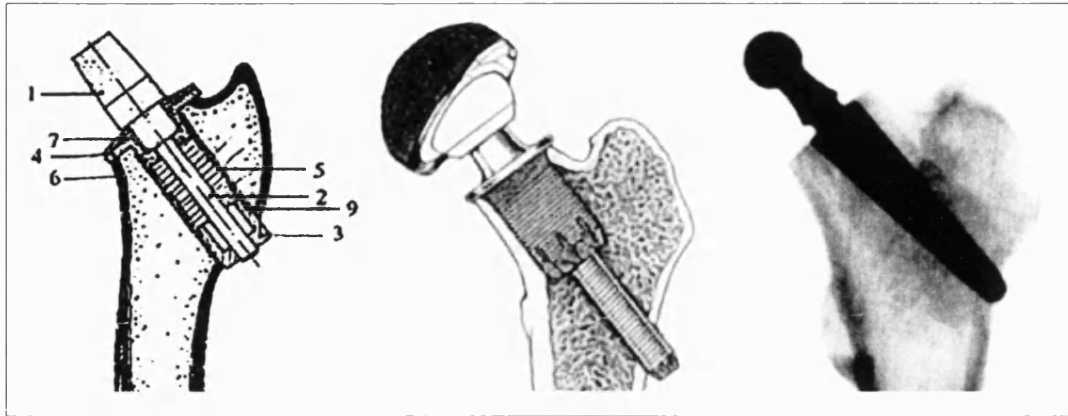


Figure 5: Images of transtrochanteric type implants (from left: Baroud¹, Gothenburg² and Willems³)

Description

A very short stemmed prosthesis which is usually straight and typically in line with the neck of the femur rather than the long axis of the bone. The load bearing stem penetrates the cancellous bone in the trochanter and can sometimes penetrate the cortical bone on the lateral side.

Structural functionality

The calcar is the primary load path with the stiff cortical bone in the medial calcar taking the majority of the compressive load. The medial cortex also acts as a fulcrum in bending between the head and the tip of the stem. Bending loads are distributed along the length of the stem to the surrounding cancellous bone and also the lateral cortical bone if the stem penetrates the cortex. Axial loads are resisted at the medial cortex.

¹ Baroud, G., Technical University of Chemnitz, Germany

² Albrektsson et al. 1998

³ Willems and Verdonschot 2002

2.4.3 Type 3: Tension anchored

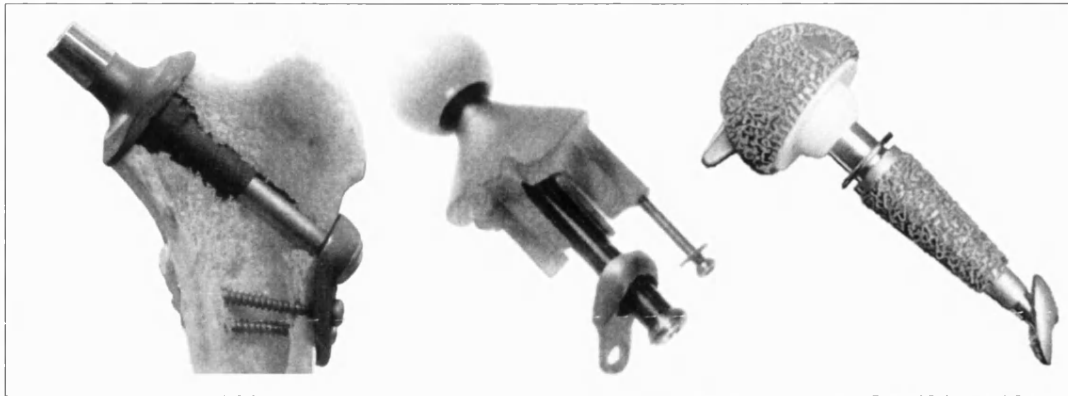


Figure 6: Images of tension anchor type implants (from left: Thrust Plate Prosthesis¹, Munting² and Cigar³)

Description

The tension anchored prosthesis is similar in design to the transtrochanteric prosthesis, but with an additional fixation to the lateral cortex. This usually comprises a plate screwed into the lateral cortex to which the transtrochanteric component is attached.

Structural functionality

Tension anchor designs primarily load the stiff cortical bone in the femoral neck in compression. Initial primary loading, interface stabilisation and positioning are usually enhanced via the tensioned component across trochanter. There is a reduction in the tendency for separation of the implant-bone interface during bending loads, due to the tension applied across the trochanter to redistribute the load to the outer lateral femoral cortex. There is enhanced anti-toggle stabilisation due to the lateral anchoring of the implant.

¹ Zimmer GmbH

² Munting and Verhelpen 1995

³ ESKA Implants

2.4.4 Type 4: Lateral abutment

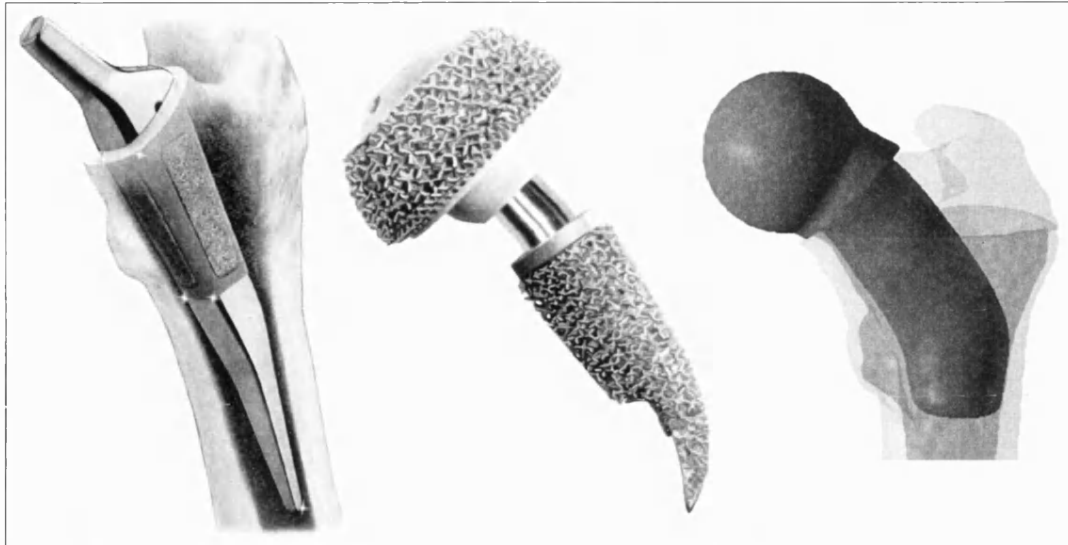


Figure 7: Images of lateral abutment type implants (from left: Mayo¹, Cut² and Non-metallic³)

Description

A short stem prosthesis which is angled with a neck in line with the neck of the femur and a tail which contacts the lateral endosteal cortical bone in the more distal region of the metaphysis.

Structural functionality

This type of implant uses three-point-bending type load transfer to bone via angulation of stem assuring medial cortical load transfer where the stem tip resists the main bending loads in the frontal plane, and the medial calcar region takes the majority of the compressive load. The stem tip acts as anti-toggle stabilisation in the frontal plane via compressive abutment against internal surface of lateral cortex. Short/medium stem assures a reasonable moment arm for stabilisation.

¹ Zimmer Inc.

² ESKA Implants

³ Zimmer GmbH

2.4.5 Type 5: Medullary stabilised

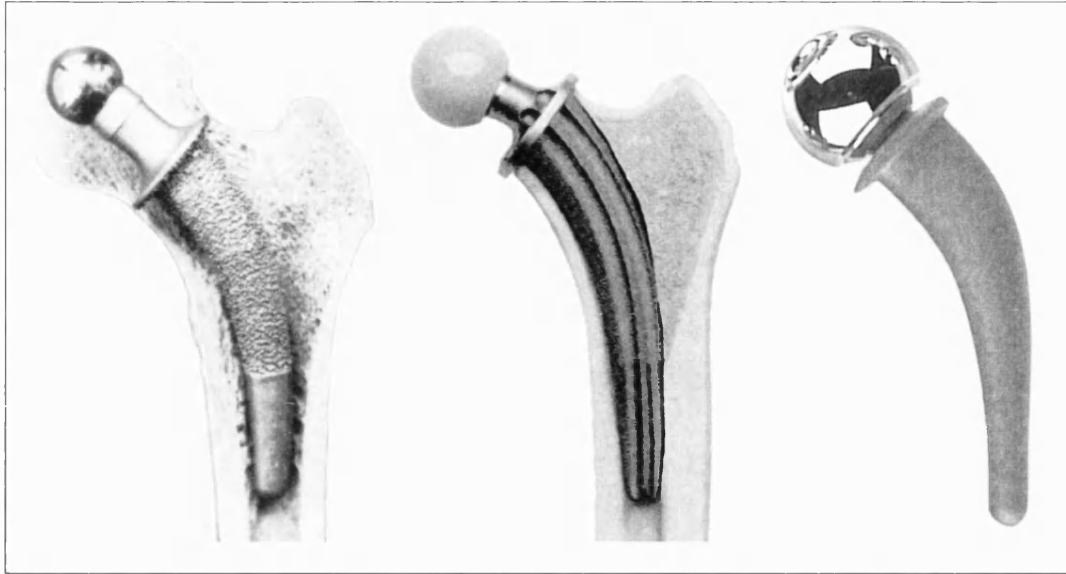


Figure 8: Images of medullary stabilised type implants (from left: Biodynamic¹, CFP² and Stelcor³)

Description

A medium-short stemmed prosthesis which follows the contour of the medial endosteal cortex and the tip rests in the start of the diaphysis of the femur.

Structural functionality

The implant typically has medial cortical contouring to promote good calcar support and preferential medial cortex loading. Distal tip stabilised in start of medullary canal providing antero-posterior and medio-lateral resistance to toggle. Thus the longer moment arm provides additional stability and more interface for load transfer. Stem length and curvature provide increased resistance to torsion.

¹ Howmedica

² LINK GmbH

³ Zimmer GmbH

2.5 History and clinical data of conservative implants

This section details further clinical evidence and previous investigations into the failure behaviour of the different groups of devices to try and identify any design related problems.

2.5.1 Resurfacing

Amstutz et al. (1998) have written a detailed review of the history of surface replacement. It was found that long-term follow-up was often disappointing for different surface replacements due to the large number of failures. These failures were attributed to wear debris related loosening. This was thought to be exacerbated by the increased wear from the larger than normal femoral head. An additional factor may be that the implant-bone interface is very close to the potential wear generating site. The early metal-metal bearing systems lost popularity due to high frictional torque which led to seizing and loosening. These failures were mainly due to manufacturing limitations, although some implants survived for 25-30 years because of minimal bone resorption.

It was thought that resurfacing may devascularise the femoral head by severing the external blood supply (Cappello et al. 1984), which may lead to osteonecrosis of the femoral head and subsequent collapse. In a 2-7 year follow-up of the Indiana resurfacing device 3/116 cases were thought to have suffered postoperative avascular necrosis followed by fracture, and 10/116 had femoral component loosening (Cappello et al. 1984). Isolated femoral component loosening was rare, suggesting wear debris may have had a role. An analysis of retrieved femoral heads from revision of resurfacing prostheses found no evidence to support general avascular-necrosis (Howie et al. 1993). When osteonecrosis was present, it was patchy and therefore attributable to loosening at the bone cement interface perhaps due to overloading of the bone, abrasion of the bone or interference with local circulation.

The resurfacing procedure was thought to be the most technically demanding conservative hip replacement procedure (Amstutz et al. 1998), and more difficult than that of stemmed type procedures. It was also said that there was a significantly lower infection rate due to lack of deep surgical penetration (Wagner 1978). Femoral neck fracture due to notching of the neck during reaming of the stump of the femoral neck was not uncommon with resurfacing operations (Murray and Meter 1982, Howie et al. 1993). Femoral neck fracture was also reported in the use of the Indiana (Cappello et al. 1978), although notching was thought not to have occurred. It is also difficult to control the uniformity of the thin cement mantle during insertion of the implant and it is thought that subsequent failure at the cement interface may cause loosening (Cappello et al. 1984). It has been found that radiographic follow-up is not a reliable predictor of failure (Cappello et al. 1984), because the implant bone interface is shielded from view by the metal implant. The operative

procedure for converting a hemi-resurfacing (no acetabular component) to a THR is relatively easy and similar in difficulty to a primary THR (Mont et al. 2001).

The survival rate of femoral surface replacements has been low relative to conventional arthroplasty (Huiskes et al. 1990). Unphysiological loading can occur in the bone directly beneath implant, leading to stress shielding and also stress concentrations around the edge of the implant leading to collapse of the femoral head or loss of stability of the implant (Huiskes et al. 1985). Cappello suggests that there is a direct correlation between an oversized femoral component and an increased risk of loosening (Cappello et al. 1984). The cup to neck ratio was determined from radiographs and there was a statistically significant correlation between this and the failed implants. Huiskes et al. (1990) evaluated several stages of loosening in a 2D FE model and found that the final disruption of the implant bone interface was probably caused by extremely high bone stresses and micromotions occurring once medial and lateral bone resorption had occurred. Different scenarios included fully bonded, loose, and various area of resorption based on histological findings. Micromotion results were consistent with findings of medial resorption progressing onto lateral resorption. The investigations suggested that the high failure rates in this type of implant are due to sensitivity of the implant design to loosening and the propagation of bone resorption and fibrous tissue formation, by the proliferation of relative motions between implant and bone. It was concluded that this type of implant can have difficulty in obtaining a stable post-loosening configuration (Huiskes et al. 1990). A more recent study conformed that the femoral surface replacement component is very sensitive to a small amount of bone resorption, causing elevation of interface stresses and motions (Vena et al. 2000). It was said by Trentani and Vaccarino (1978) that the femoral epiphysis did not undergo any changes so long as the femoral prosthesis was fixed to it but that, as soon as the prosthesis loosened, progressive damage occurred in the neck-epiphysis complex, which reduced its dimensions concentrically.

2.5.2 Transtrochanteric

The transtrochanteric component in this type of device can be subjected to significant bending/tension loads. This behaviour was demonstrated by the Judet prosthesis, which tended to break without a metal rod reinforcement (Judet and Judet 1950). However, it should be noted that the Judet prosthesis was relatively weak, compared to modern implant materials, as it was typically made from acrylic.

Haboush (1953) carried out a theoretical analysis of some different transtrochanteric type designs and predicted that high stresses would occur at the neck pivot point and also the stem tip. This seems to agree with the results of another study which showed that the probability of interface failure can be higher in shorter stemmed implants van Rietbergen and Huiskes (2001). Whether adequate stability can be achieved with such short implants is obviously in question, as demonstrated by a stability study of a novel transtrochanteric

device (Willems and Verdonshot 2002), which acknowledged that the load carrying surface area was reduced in comparison to conventional implants. It was also noted in the study that the penetration of the lateral cortex may weaken the bone.

2.5.3 Tension anchor

In a clinical trial of the Munting prosthesis, after a mean follow-up of 4 years, it was said that no loosening migration or fracture of the bone or screw failures had occurred in the cases which were well-inserted (Munting and Verhelpen 1995). However, other failures did occur, and it was suggested that implant positioning may have affected load transfer and caused bad results (Munting and Verhelpen 1995, Munting et al. 1997). A statistical analysis showed a strong correlation between implant position and resorption of the calcar femoral, as well as varus implantation and clinical failure (Munting et al. 1994). However, a subsequent investigation showed no significant difference in stability between different angles of implantation (Munting and Verhelpen 1995). Clinical results of the Thrust Plate Prosthesis showed no difference in clinical results from different implantation angles (Fink et al. 2000).

Significant loads can be transmitted through the tensioning components. This was demonstrated by screw (5/100) and lateral plate (5/100) fractures in the Jaenichen-Collinson prosthesis (Peterson 1950). Another early design of prosthesis based on a lag-screw (McKee 1970) fractured the neck of the femoral plate after 3 years in one of 3 patients treated (the other 2 became loose with 1 year). This design was constructed of stainless steel and did not have any collar support, so failure may have been inevitable. This shows the way in which load can be transferred to the laterally fixed area if collar support is not present. A brief investigation into potential problems that may occur with a lateral plate design by Haboush (1953) indicated that loading problems may occur due to improper seating of the lateral plate. The lateral zone is subject to tensile forces in the case of incomplete collar contact leading to pivoting at the neck and excessive tension on the lateral plate. The pressure on the pivoting neck in such designs was illustrated by bone failure in the calcar region during in vitro testing of an implant (Munting and Verhelpen 1995). Fatigue fracture of the implant's main tensioning bolt also occurred several times in developmental testing phase (Munting and Verhelpen 1995). The bolt diameter was subsequently increased from 5 to 8 mm.

2.5.4 Lateral abutment

Behaviour of this type of prosthesis was demonstrated in a follow-up of 159 Mayo prostheses over a mean of 6.2 years (Morrey et al. 2000). The Mayo had more cases of proximal osteolysis (11/159) than distal osteolysis (3/159), of which 9/14 required revision. Subsidence was found to be relatively common in this type of implant with 20/159 having measurable subsidence. The subsidence was sometimes accompanied by

loosening and required revision. It was admitted that early mechanical loosening (3/159) due to inadequate fixation was greater than that observed with some other designs (Morrey et al. 2000). A similar study was published at 2 years follow-up (Morrey 1998). It was said that incorrect placement/sizing of the Mayo may have led to subsidence failure of the implant (Morrey 1989).

In a FE study of the Mayo prosthesis (Huiskes et al. 1986), the fixation stability of the prosthesis was thought to be doubtful if calcar resorption occurred. An FE model of the implant, indicated minimal subsidence due to the strong taper, but with considerable increases in interface compression. It was also thought that there was a risk of failure of the distal part of the stem, owing to the significant loading through the distal tip (Huiskes et al. 1986) due to its abutment against the lateral cortex. A study of the CUT prosthesis in dogs indicated that varus migration was expected in this type of prosthesis (Oldenburg et al. 2002) and concern was expressed over the increased load and tension forces created at the lateral contact area and their effect on the long-term outcome.

2.5.5 Medullary stabilised

An early design of the Biodynamic prosthesis showed signs of stress bypass with calcar and trochanter resorption and also sclerosis of the bone in the distal third (Pipino and Calderale 1987). Experimental investigations of different sizes of Biodynamic prosthesis implanted into composite and cadaveric femurs showed that over-sizing of implant could cause significant alterations in ideal load transfer (Fagan et al. 1996) reducing significantly the strains in the proximal region. Loosening and incongruity on the lateral side of the stem had also been found in 3/17 cases of the Biodynamic (Pipino and Molfetta 1987). It was noted that stems have also tended to migrate by sinking or going into varus, causing poor results (Pipino and Molfetta 1987).

2.5.6 Summary

Generally it was found that there was not a significant amount of evidence of long-term, fatigue type problems evident for conservative designs. This may be because many devices were plagued by early problems and thus never stayed implanted long enough. Most reported mechanical failures related to prototype testing, and much of the older evidence is anecdotal and not detailed. From the evidence available, some patterns were discernible. The shorter implants (resurfacing and transtrochanteric) seemed to suffer more from stability based problems, whereas it appeared that the longer implants (medullary stabilised and lateral abutment) were affected by stress bypass phenomena and subsidence. The mid-length implants (tension anchor) seemed to suffer from the effects of significant loading through the distal anchorage point, especially when collar loading was absent.

2.6 Analysis of implant groups – effect of implant design

2.6.1 Aim

The aim of this investigation was to carry out FE analysis of the different groups of implants. This was done to enable a better understanding of the behaviour of the different types of implant and investigate if this was indicative of clinical behaviour.

2.6.2 Investigation outline

Finite element analysis was used as a tool to explore potential behaviour of the bone implant construct. Models of typical implants from the different classification groups of conservative implant were studied regarding their effect on a well validated bone model. Standard length stems were included in the analysis for comparative purposes. The study also examined the effect of such features as the collar and their relative importance in the different designs.

2.6.3 Method

2.6.3.1 Geometry

The bone model used was created by other researchers for a previous investigation (Taylor et al. 2002) from computer tomography (CT) scan data from a cadaveric bone. The bone was of typical size with no evidence of joint disease. Semi-automatic edge detection software had been used to create periosteal contours from the CT data. These had then been used to create a solid model in Unigraphics V15 (Figure 9).



Figure 9: Cadaveric bone model

The plan for the implant models was to simplify each type of design to its key geometric features. This had the following effects:

- a) reduced the effect of implant-specific features
- b) enabled a more generic view of the implants within each group
- c) enabled better comparison between implant groups

Implant modelling was carried out by the author in Unigraphics V18 (Figure 10). Stem geometries were created from a linear swept blend along an implant centreline. The implants started at a common initial cross-section at the resection level tapering to a common tip cross-section. The changes between implants were only in length and curvature. The effect of the common collar was optional in the FE investigation (i.e. collar and no-collar were both possible). Not only was the analysis regarding effect of collar removal related to geometric design, but it could also indicate potential variations caused by loss of collar contact, due to surgical errors or possible bone remodelling effects.



Figure 10: Geometries of the simplified implant groups (from left: Resurfacing, Transtrochanteric, Tension-anchor, Lateral abutment, Medullary stabilised and Traditional stem)

The resection level and implant placement was based on typical characteristics of implanted metaphyseal implants. The implant cross-section at the resection level was derived from an average shape taken of several different implants giving an elliptical shape. The size was judged based on the other implants' dimensions giving a reasonable level of cortical contact. This was governed by bone shape and fit within the neck to provides rotational stability and cortical loading. For the tip cross-section the average was taken of several different implants tip leading to 10mm circular cross-section. It was noted that nominal sizing of this dimension is likely to be governed by the local morphology of bone (i.e. medullary canal diameter). The chosen tip geometry balanced these constraints. No press-fit was included in the model, as there was no information available as to

whether this would be retained long term. The collar and neck geometry were consistent for all implant geometries where it was required. For the resurfacing implant the cup was placed coaxial with the neck, with depth of insertion commensurate with required leg length. Resurfacing components are commonly cemented, as this gives much more versatility in treating patients with misshapen femoral heads, large bone cysts or avascular necrosis (McMinn et al. 1996). The resurfacing design was hence modelled with cement, with a layer of cement with an average depth of approximately 1mm.

2.6.3.2 Meshing

The geometry from Unigraphics was converted to a suitable format to be imported by the ABAQUS FE software. Virtual topology (ABAQUS/CAE) was then carried out manually on the model to remove small artefacts and slivers which occurred during the transformation process. Unchecked, these would unduly affect the mesh density in local areas, and could also lead to degenerate element shapes.

For computational efficiency the whole length of the femur was not analysed since the more distal region, towards the condyles, was not of significant interest. Thus the model was truncated in the middle of the femur, giving enough clearance between the most distal part of the longest implant and the required boundary conditions. Boundary conditions were implemented at the distal end of the model, such that nodes on the truncated surface were restrained. An investigation by Polgár et al. (2003) found altered strain pattern due to fixation methods, at a distances less than the diaphysis of the femur. The boundary conditions were considered to be suitable since they exceeded this level in the area of interest, having reasonable (50mm) clearance from the tip of the longest model.

In a study of the accuracy and efficiency of different element types Polgár et al. (2001) found that linear tetrahedral elements should be avoided and quadratic tetrahedral elements ought to be chosen for the purposes of FEA of the human femur, as they provide a more realistic description of the behaviour of the structure. It is noted in this paper that a particular mesh can be considered sufficiently refined when the strain energy does not increase significantly with subsequent mesh refinements. Although this holds true for homogenous linear elastic models used in Polgár's study, for the current investigation this criterion is not applicable since the materials properties vary on an element by element basis.

A study by Perillo-Marcone et al. (2003) assessed the effect of mesh density on material property discretisation and the resulting influence on the predicted stress distribution. The study showed that the assigned material properties could be highly dependent on the element size and the element size should be similar to the CT scan size. However, for the continuum assumption to remain valid, elements should be at least 1-2mm in size (Perillo-Marcone et al. 2003).

The geometry of the femur was complex and required automatic meshing routines for efficient mesh generation. From a practical point of view tetrahedral elements were the best choice for automatic mesh generation when a solid model is available (Viceconti et al. 1998b). Linear tetrahedral elements are overly stiff and require very fine meshes to produce accurate results (ABAQUS 2002). In general quadratic tetrahedra are better as they can achieve a better geometric fit and they capture stress concentrations more effectively. But they are not suitable for contact analyses. This is apparent in uniform pressure situations where significantly different contact forces can arise at the corner and midside nodes. Modified tetrahedral elements are designed to alleviate these shortcomings. Modified tetrahedral elements are recommended for use in contact problems, because the contact forces are consistent with the direction of contact giving rise to uniform contact pressures in uniform pressure situations (ABAQUS 2002). They are also robust during finite deformation.

After mesh generation, mesh quality checks were carried out. Checks were carried out using ABAQUS/CAE's mesh analysis tools. These monitored the shapes of the elements and warned of badly shaped elements which could cause erroneous results or failed analyses. Shape anomalies were minimised and failed analysis checks were eliminated completely, by adjustment of mesh seeds in difficult areas.

To investigate the effect of mesh refinement level, a sensitivity study was carried out on the unimplanted model. Three levels of mesh refinement were investigated with average element edge length of 2.5, 3.0, and 4.0mm. The peak von-Mises stress was noted for each model (Table 3). This occurred on the medial surface of the femoral neck in all cases. It was interesting to note that the region of highest stresses during the loading was coincident with the ridge which is apparent from the neck to the origin of the psoas major, which may be indicative of the relationship between form and function.

<i>Approximate mesh size (mm)</i>	<i>Peak von-Mises stress at neck surface (MPa)</i>
4	49.8
3	60.7
2.5	64.1

Table 3: Mesh sensitivity study

Resource limitations meant that a mesh sensitivity of less than 3mm was impractical for cases involving contact. Although the mesh size was not achieving convergence it was felt that the results would still be useful for comparative purposes, since the meshes had similar levels of refinement.

2.6.3.3 Materials

Implant materials

Implant materials were typically modelled as Titanium alloy ($E = 110\text{GPa}$, $\nu = 0.33$). This is a commonly used material for implants of all types as it can integrate well into the host bone. The exception to this material was the material used for resurfacing prostheses. The bearing surface is typically Cobalt Chrome ($E = 220\text{GPa}$, $\nu = 0.3$), as it has low wear and does not weld (unlike Titanium alloy). The pin in the resurfacing design was also modelled as Cobalt Chrome. This was similar to the Durom design and was considered appropriate for the generic case – as these pins are typically intended for initial alignment only and not long term integration into the host bone and subsequent load bearing.

Cement materials

It is known that residual stresses can occur during the curing process/cement expansion, but cement creep also decreases cement stresses; this is therefore a complex issue dependent on boundary conditions (Nuno and Amabili 2002). For this investigation such complex issues were neglected and it was assumed that the cement properties are $E = 2.6\text{GPa}$ and $\nu = 0.3$. These were the same as used in a previous unpublished investigation using this cement, and similar to those values used by Nuno and Amabili (2002).

Bone materials

The bone's materials properties were assigned using a programmed method developed by other researchers previously (Taylor et al. 2002). The complex trabecular structure of bone was represented with a continuum description, whereby each element was assigned a density by the programme based on CT scan data, and appropriate orthogonal material properties and directions. The material assignment programme used the meshed model of the femur as a starting point. A series of co-ordinate frames were created which rotated around the longitudinal bone axis. In the head/neck region of the femur, the coordinate frame was rotated by the neck angle. The programme calculated the mid-point of each element and searched the femur's CT data files for a voxel including these coordinates. That voxel's Hounsfield Unit (HU) value was then assigned to the element. The HU values were grouped in intervals of 50HU to limit the number of material cards to a manageable amount. Density values were calculated from the HU values using the method developed by Taylor et al. (2002). Material properties were calculated from the density using the relationship shown below (where E =Young's modulus, and ρ =bone density). Maximum values were obtained from ultrasound measurements from bone specimens.

$$E = E_{\max} \frac{\rho^2}{\rho_{\max}^2}$$

The output of the code was used to create an Abaqus input deck with appropriate material properties and orientations for the elements. The properties of the bone produced in this manner were well validated in a previous study (Taylor et al. 2002).

2.6.3.4 Interface conditions

Initially it was assumed that the implant and bone were fully tied – as this was less computationally expensive. But Keaveny and Bartel (1993) indicated that using fully bonded surfaces could lead to numerical artefacts (high shear stress with low normal stress). Indeed this was found to be the case as shear stresses indicated isolated nodal peaks of interface shear stresses. As this was clearly not a realistic representation, the models were modified to include a friction model, which would be representative of the case immediately after implantation without boney integration.

The friction model for the implant/bone interface was 0.4 friction coefficient with an elastic limit of 20 μ m to account for non-linear interface behaviour. This was based on the investigations of Shirazi-Adl et al. (1993) who carried out experimental testing to determine the friction characteristics of the metal to bone interface for different surface structures. They investigated the effect of cyclic loading and found that there was typically a 14% decrease in the friction coefficient after loading at 1Hz. It was also found that relative displacements of 50-400 μ m occurred before peak resistive force occurred. Another study by Kuiper and Huiskes (1996) used a friction coefficient of 0.4 which corresponded to experimentally determined coefficients of friction for wet bone on smooth titanium.

It is known that PMMA does not form chemical bonds with implant (or bone) so bonding strength is dependent upon mechanical interlocking of the surfaces (Lucksanasombool et al. 2002, Nuno and Amabili 2002). For the purpose of this investigation it was assumed that the cement was well fixed to the bone. Hence, tied contact was assumed between bone and cement, but no cement penetration was physically modelled. Implant to cement was modelled with 0.3 coulomb friction. This was based on an investigation by Mann et al. (1991) who carried out testing and FE on the interface between Ti alloy and PMMA and found that coulomb friction with a value of 0.3 characterised the interface well.

The non-linear solver was used for the contact simulation. The small-sliding formulation was used as the relative movement was expected to be within the range suitable for this and also this is far less computationally intensive than large sliding formulations. Slave surface interfaces were initially adjusted (without strain) so that contacting surfaces were initially perfectly in contact. Automatic overclosure and separation tolerances were used to speed up cases where contact was slow to establish numerical stability.

2.6.3.5 Loading

Muscle forces

Muscle forces (Table 4) were obtained from a study by Duda et al. (1998) and based on the 45% swing phase of the gait cycle (peak contact force). These muscle forces were based on a standardised femur which has very little anteversion or anterior bow. The bone model used in this investigation was of similar size to the standardised femur, so no scaling of the muscle forces was used. Muscle origins and vectors were first reflected to account for the opposite leggedness of the model. Orientation of the co-ordinate axes in Duda et al.'s model was based on alignment with the femoral condyles, but due to the anteversion and anterior bow present in the cadaveric bone, this alignment method would have caused misalignment in many of the proximal muscle groups. Hence the muscle vectors were reoriented manually, until alignment with the majority of the proximal muscle origins and surface geometry was achieved. Since only the top half of the cadaveric bone was being used in this investigation and due to the proximal anchorage position of the implants being investigated it was felt that this alternative orientation was preferable. Muscle vectors were also transferred manually to the new vectors, in the global co-ordinate system of the cadaveric model for compatibility.

<i>Muscle force</i>	<i>FORCE VECTOR (N)</i>			<i>Magnitude (N)</i>
	<i>x</i>	<i>y</i>	<i>z</i>	
<i>Hip joint load</i>	832.4	-391.9	-1987.5	2190.2
<i>Gluteus maximus 1</i>	-79.3	37.3	97.7	131.2
<i>Gluteus medius</i>	-155.6	111.3	238.9	306.1
<i>Gluteus minimus</i>	-162.1	200.8	119.4	284.3
<i>Tensor fasciae latae</i>	-33.9	60.6	-32.9	76.8
<i>Piriformis</i>	-74.4	9.1	70.5	102.9
<i>Gemelli</i>	-32.8	8.0	39.2	51.7
<i>Illiopsoas</i>	-42.4	110.1	128.1	174.2
<i>Gluteus maximus 2</i>	-55.2	0.8	72.3	91.0
<i>Vastus intermedius</i>	4.6	14.9	-61.0	62.9
<i>Vastus lateralis</i>	-21.3	95.0	-206.3	228.1
<i>Biceps</i>	6.5	-6.9	-91.5	92.0

Table 4: Table of forces reoriented from Duda et al. (1998)

It was interesting to note, that after the alignment procedure the hip contact force vector seemed to be aligned with a plane through the centre of the neck of the femur. Muscle forces were distributed uniformly over the muscle areas, by dividing the force equally between all nodes. It is realised that variations in mesh density may have caused some

non-uniformity in loading, but this was thought tenable as the mesh density was reasonably uniform. Where only part of the muscle was accounted for in the model due to the truncation of the femur model, the muscle loads were adjusted proportionally with the area remaining. Where a muscle's magnitude was considered insignificant ($<1\%$ of the contact force) it was not applied to the model. The implants under investigation were typically modelled without the ball-head for reasons of computational efficiency and the hip load was applied directly to the edge of the taper. This was considered to be a reasonable approximation for this comparative study.

Load distributions

Previous investigations have used either single point attachments for muscles or distributed the load over arbitrary small groups of nodes to try and reduce artefacts caused by the high stress gradients in muscle attachment areas (Taylor et al. 2004). For this investigation it was proposed to simulate the effect of muscle loading in a more realistic way to avoid numerical artefacts and provide better load distribution. Rather than arbitrary attachment regions, it was decided to distribute the loads over physically meaningful muscle attachment areas (Figure 11). As the bone model used in the investigation was based on a previous cadaveric study, no information was available regarding the specific muscle attachment areas for the bone in question. An appropriate muscle map for all the muscles on the femur was determined using information on shape and position from McMinn and Hutchings (1985) and the Muscle Standardised Femur (Viceconti et al. 2003) together with bone specific inferences from the geometry of the bone in question.



Figure 11: Muscle attachments areas

Contact force

The most significant load on the model was the hip contact force. It was thought that a realistic joint contact area would be more appropriate than a point load. For the natural joint this was derived from an in vitro study of the contact stress distributions in the human hip (Brown and Shaw 1982). The regression value was taken from the study's series-wide spatial mean contact stress versus applied load correlation ($0.00108 \text{ MPa}\cdot\text{N}^{-1}$). It was considered that a mean contact stress would be easier to implement than the actual state of spatial variation in the contact stress, but still provide a reasonable representation of the loading. Due to the nature of the joint the contact pressure was assumed to apply uniformly over a circular area.

Joint load	=	2190.2 N
Mean joint contact pressure	=	$0.00108 \times 2190.2 \text{ MPa}$
	=	2.365 MPa
Contact Area	=	$2190.2/2.365$
	=	926 mm ² (equiv. circle radius 17mm)

The projected circular contact area was initially centred on the load vector origin derived previously. The contact area was then slightly adjusted in position, perpendicular to the original load vector, so that it did not intrude upon the fovea capitus and that the load vector was coincident with the surface normal at the joint contact centroid. The loads were applied to nodes within the contact area of the meshed model. Loads were applied in the direction of the required vector, rather than as a pressure normal to the surface, as this would provide more control over the load due to the non-uniform surface of the femoral head.

The contact area for the natural joint is obviously going to be far larger than any man-made replacement, due to the superior conformance and load distribution properties of the cartilaginous layers. At the time of investigation no contact pressure/area investigations were available for the type or size of resurfacing replacements under investigation. The McKee-Farrar implant bearing has been cited as being similar in nature to that of hip resurfacing implants (Schmalzried et al. 1996). An appropriate contact area for the resurfacing prostheses was derived from predictions from a theoretical contact mechanics model (Jin 2000). Data was used from a typical McKee-Farrar metal-on-metal replacement, under a load of 2500N, which had approximately the same diameter as the resurfacing implant under investigation (42mm cf. 44mm). This investigation indicated a circular contact area with a radius of approximately 9mm. It was realised that actual materials properties and geometry would be different from the McKee-Farrar representation. Notwithstanding this, modern implants would achieve better tolerances, and a resurfacing prosthesis may be more flexible than the solid head of the McKee-Farrar. It was reasoned that this representation value gave a worst case indication of the

contact pressure. It was also considered that this worst case contact pressure could lead to a worst case loading distribution on the underlying bone, due to non-uniformity of loading. It was thought that in the absence of any other information this would provide a reasonable approximation for the case under investigation.

2.6.4 Results

To assess the effect of implant design, three major areas were investigated. Firstly, the effect on the implant itself was investigated by the stresses generated within the implant, which could indicate whether an implant was prone to fatigue failure. A common indicator for a material performance in fatigue is the von Mises stress. The values obtained by the analysis were compared to the typical fatigue limits of the materials in question.

Secondly, the effect of the design on the host bone was of interest, and any possible biological reactions which could be induced due to the change in loading of the host bone. An area of concern was the potential change in bone density. Bone remodelling indications were assessed by comparing the levels of Strain Energy Density (SED) in the bone before and after the implantation of the device. Increases in SED would be indicative of a potential to increase in bone density, and decreases in SED would be indicative of the potential for bone density to be lost. SED is thought to be related to bone remodelling as it has been used in bone remodelling simulations as the driving signal which determines whether bone remodelling occurs (van Rietbergen et al. 1992). To show the remodelling indications upon loading of the implants, similar cross-sections were shown for comparison between the natural unimplanted bone, and the implanted versions. Due to the different meshes used in the implanted and unimplanted models it was impractical to try and calculate the differences automatically. Scales were chosen to try and highlight the changes which were occurring, although spatial variations in SED levels sometimes made this difficult.

Finally the interaction between implant and bone was of interest, as the stability of the implant/bone construct can influence the success of the hip replacement. Stability was assessed by examining the levels of micromotion which were developed at the interface between implant and bone. Micromotion values above 150µm are known to cause interface tissue to develop, and values below 20µm are required for bone integration to occur (Jasty et al. 1997).

2.6.4.1 Implant stresses

The peak von Mises stress for each type of implant are shown in Figure 12. Peak stresses typically occurred in the neck region. All stresses were well below the fatigue limits of the materials in question. The theoretical endurance limit is approximately 610MPa for Titanium alloys and approximately 700MPa for Cobalt Chrome alloys.

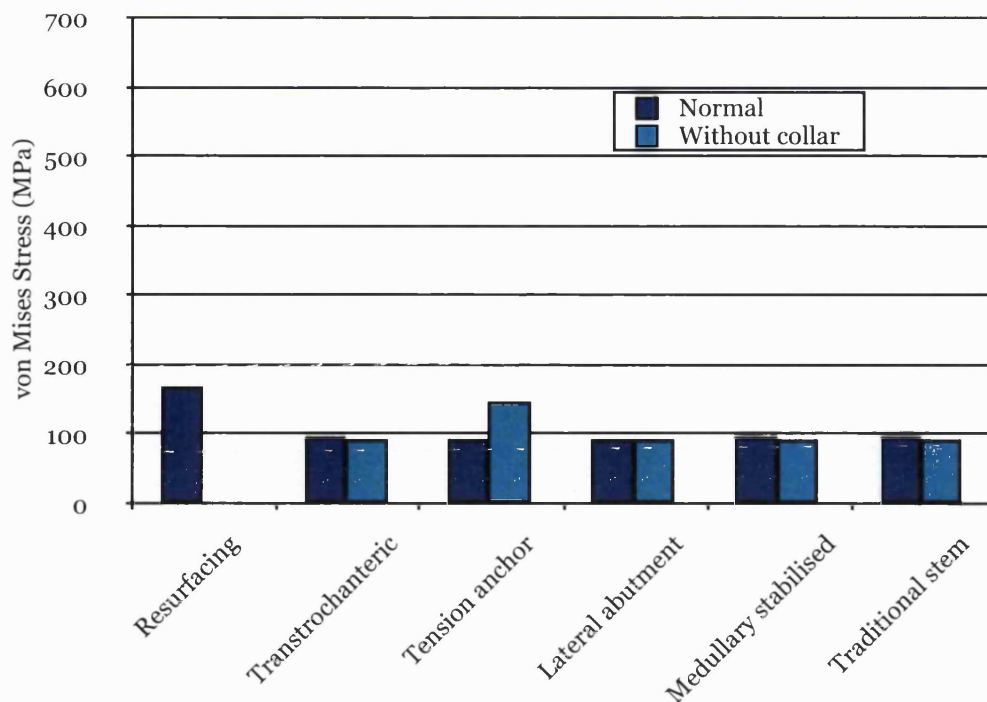


Figure 12: Peak implant stresses for different implant designs

2.6.4.2 Bone remodelling indications

The resurfacing prosthesis (Figure 13) showed a significant reduction in SED in the superior region, under the cup. A sample point in this area showed a fall of 84% in SED. There were also smaller areas of SED reduction near the tip of the implant. Over the rest of the bone, there appeared to be no significant change in behaviour from the natural model.

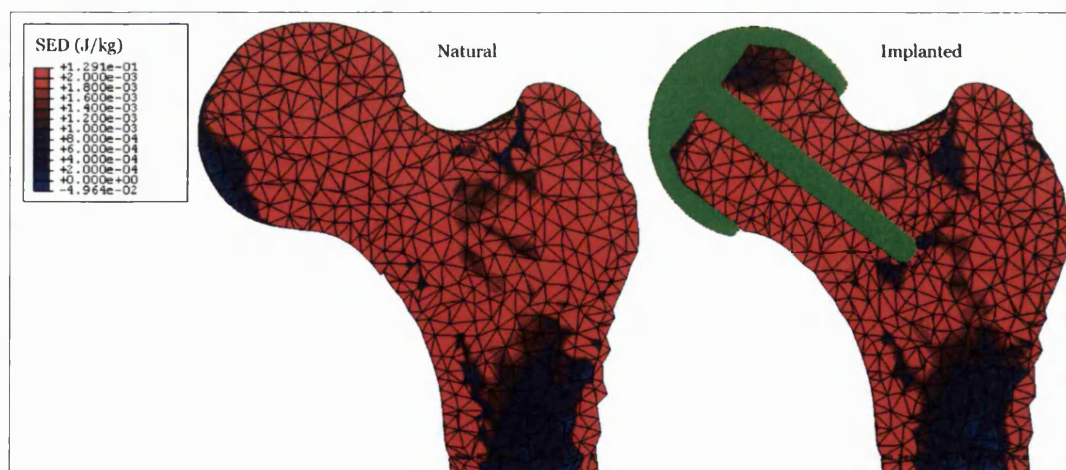


Figure 13: SED levels showing the effect of the Resurfacing design

The transtrochanteric prosthesis showed increases in SED near its distal end (Figure 14). This was increased further without the collar. There was no noticeable reduction in SED in the proximo-medial region, with or without a collar.

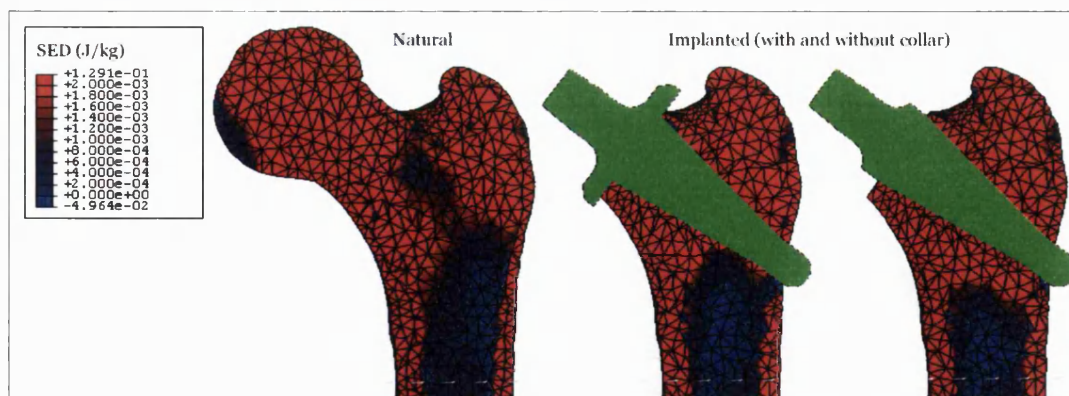


Figure 14: SED levels showing the effect of the Transtrochanteric designs

The tension anchor showed similar behaviour to the transtrochanteric implant with some SED reduction immediately adjacent to the lateral plate (Figure 15).

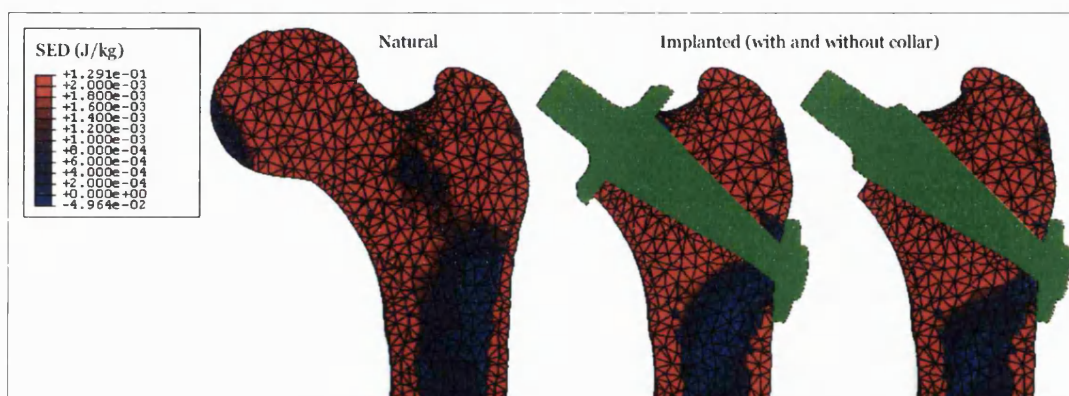


Figure 15: SED levels showing the effect of the Tension anchor designs

The lateral abutment implant showed some SED increases in the midstem region, and increases at the distal tip upon removal of the collar (Figure 16).

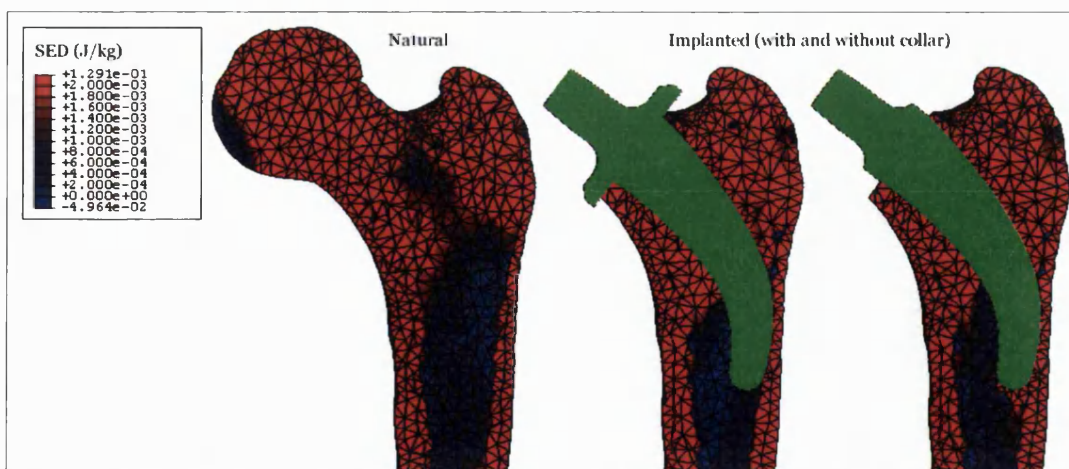


Figure 16: SED levels showing the effect of the Lateral abutment designs

The medullary stabilised implant showed increases in SED at the distal tip (Figure 17). Removal of the collar indicated further increases in SED at the distal tip and also some losses in the proximo-medial areas. The traditional stem showed increases in SED at the distal tip (Figure 18). The removal of the collar increased the SED significantly at the distal tip with notable reduction in SED in the proximo-medial region.

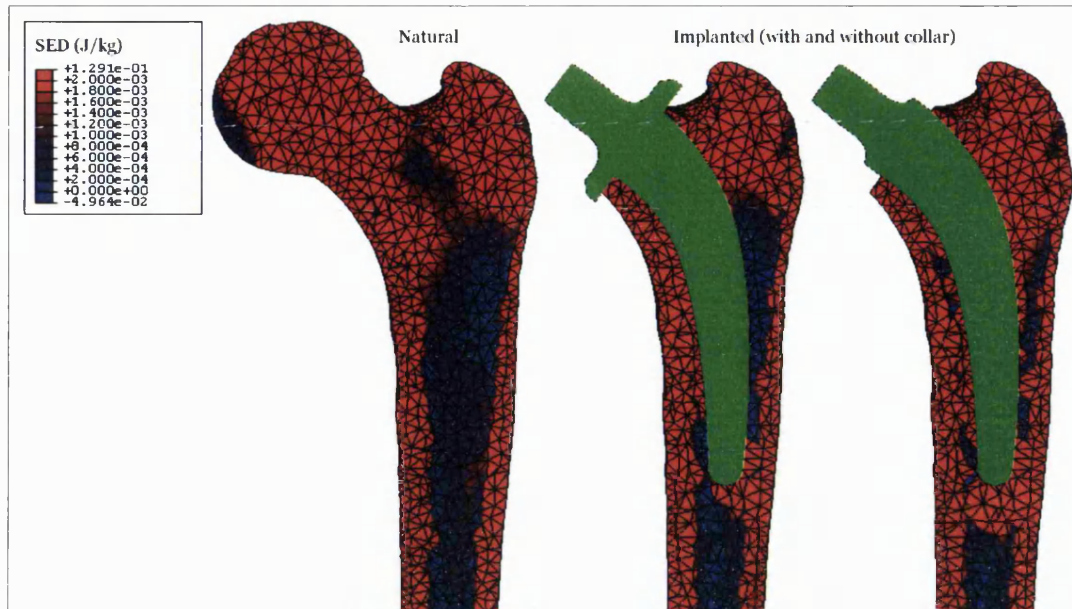


Figure 17: SED levels showing the effect of the Medullary stabilised designs

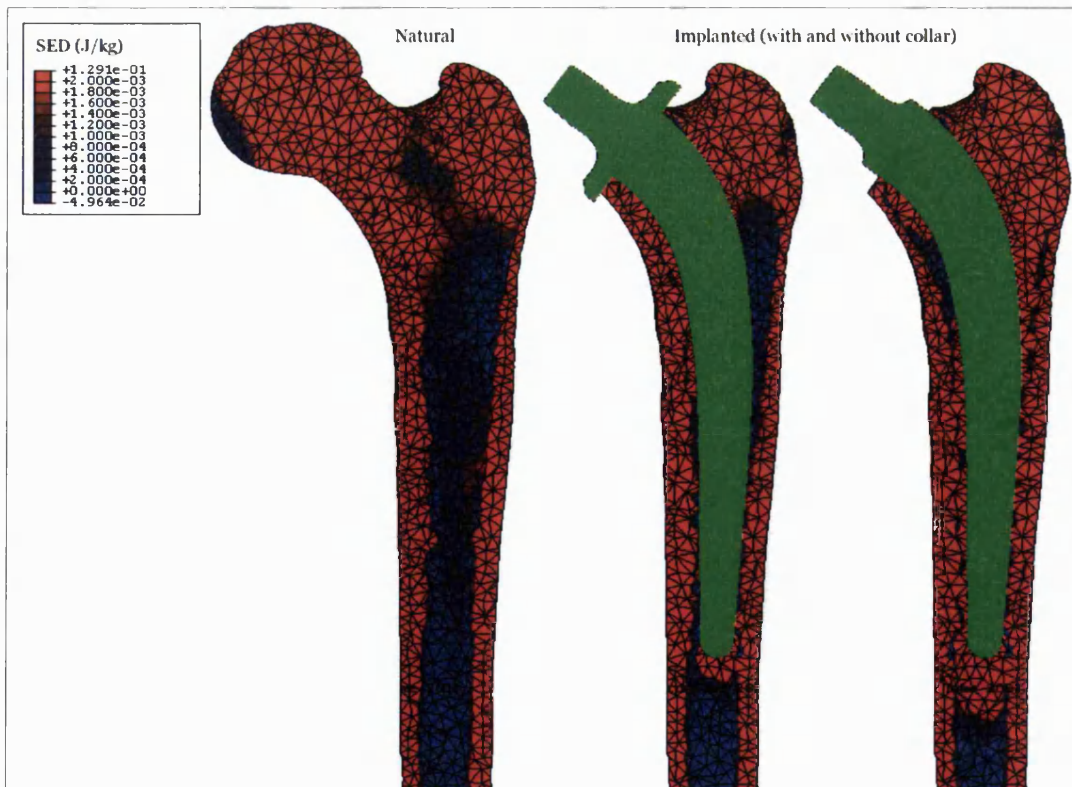


Figure 18: SED levels showing the effect of the Traditional stem designs

To highlight the overall effect of implant design on the SED a point was chosen in the proximo-medial calcar area for further investigations. This region was of particular interest, as bone mass is typically lost in this area in the clinical situation. The same point was investigated for all implants. The results for this point (Figure 19) showed that the shorter implants have increased SED in the proximo-medial area over the longer implants. It can also be seen that removal of the collar led to a much steeper decline in SED with increasing implant length.

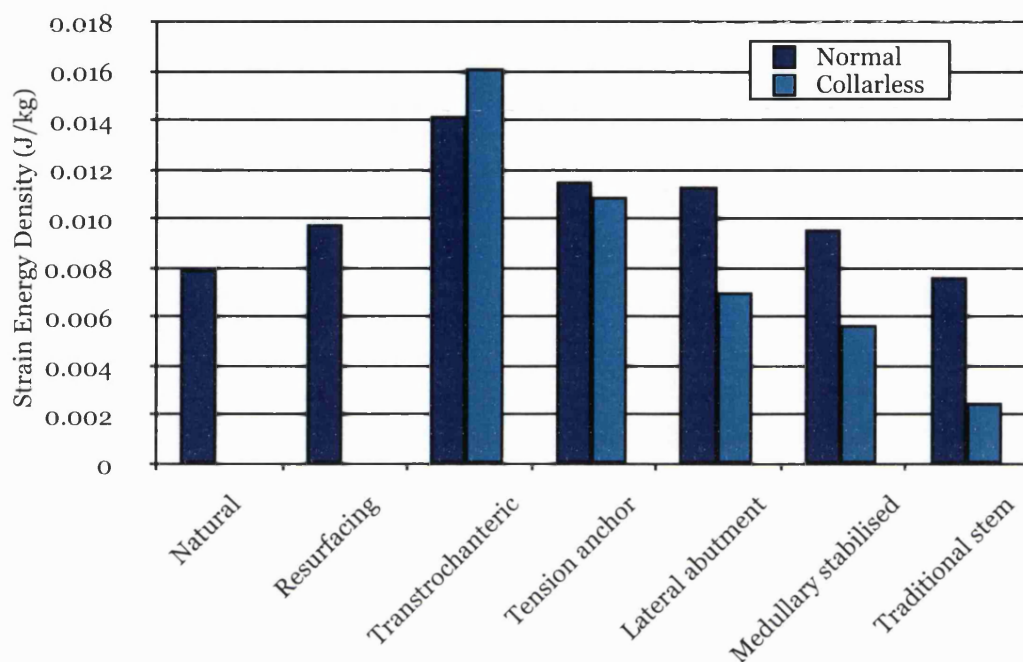


Figure 19: Bone remodelling indicators in the medial neck

2.6.4.3 Implant micromotion

To investigate the stability of the implants the peak micromotion at the interface between implant and bone was noted (Figure 20). No general trend with implant length was seen, although it is noted that the micromotions of the implants without collars are significantly higher than those with the collar present.

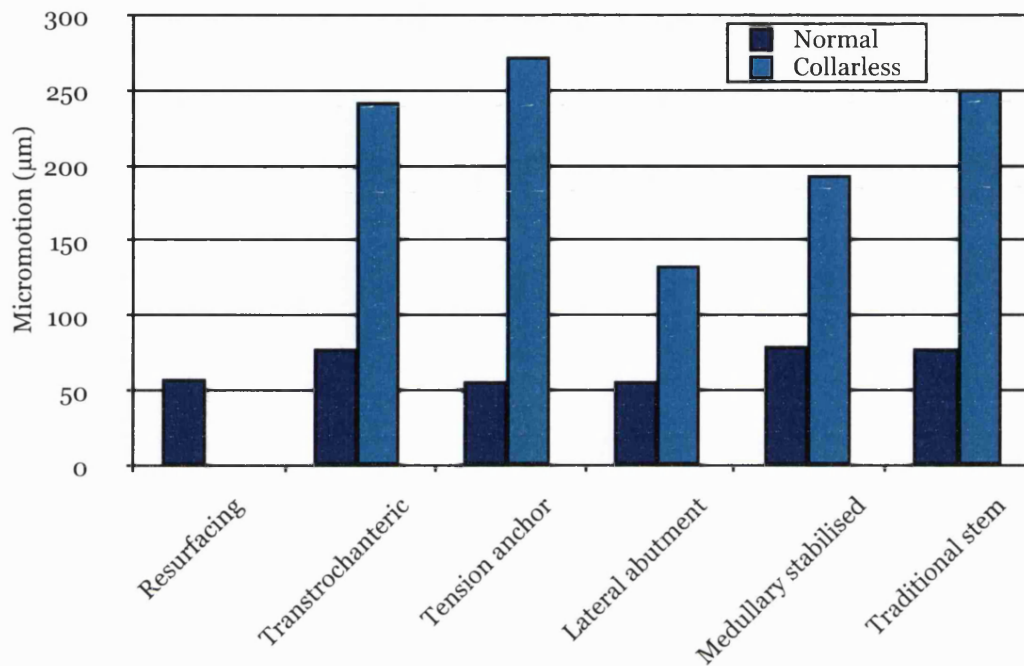


Figure 20: Peak micromotion at implant/bone interface

2.6.5 Summary of results

There was no evidence of any particular type of implant prone to fatigue. The peak implant stresses were typically seen in the neck region of the stems. There did not appear to be any effect on the implant stresses due to implant length. All implants indicated peak stresses well below the fatigue limit of Titanium (610MPa). The loss of collar support had a negligible effect on the implant stresses, except for the Tension anchor type of prosthesis where the increased loading on the lateral plate caused elevated stresses.

There appears to be a relationship between the amount of proximal load transfer and the length of stem. The SED results show that the shorter implants increased the amount of loading through the proximal region, over the physiological situation. It is expected that this could lead to increased bone density in the proximal region for such implants.

The longer stems demonstrated increased levels of SED at the distal tip of the implant. This is indicative of the stress bypass which is thought to occur in such stems. This stress bypass could lead to increased bone density distally, and subsequent distal jamming which may lead to the worst case cantilever bending on the stem. From the results it appears that

the longer the stem, the greater the stress bypass becomes. The shorter implants appeared to be at a lower risk from the stress bypass phenomenon.

The loss of collar support appears to have notable consequences on the SED, especially for the longer stemmed implants, as this causes an even larger amount of load transfer to the distal end. This would in turn increase the risk of detrimental bone remodelling.

The micromotion results do not show a significant trend dependent on stem length. It is notable that the loss of collar support universally led to a significant loss of stability, indicated by the micromotions of above 150µm. It is thought that the fit and fill of the implant may have a significant effect on implant stability, but this was not considered in this study.

Resurfacing designs were found not to impact significantly on the overall physiological bone stress although there is concern over localised reductions in bone stress immediately under the cup in the superior region. This could lead to bone loss and subsequent destabilisation of the implant.

2.6.6 Discussion

There was no significant evidence of fatigue from most of the different implants, and this is in general agreement with the lack of clinical evidence of fatigue failure in conservative implants. However, it should be noted that there was limited clinical evidence for the types of implants investigated. Fatigue indications were slightly elevated for the Tension anchor type device, when collar support was lost. This caused elevated stresses in the lateral fixation component. This is in line with the anecdotal evidence of some implant failures in this area during prototype testing (Munting and Verhelpen 1995), although this evidence is only anecdotal in nature.

The indications of bone remodelling suggested that the longer implants were more prone to the effects of stress bypass, where there is an increased risk of proximal bone loss and distal bone sclerosis. In a combined FE and mechanical testing study by Tanner et al. (1995) it was also found that decreasing stem length tended to increase the load transfer to the calcar region. Thus it is expected that the shorter implants are not as prone to the worst case cantilever bending which arises from the loss of proximal bone support typically seen from long stems.

There was no discernible trend with stem length for the stability of the implants. However, the loss of collar support universally led to a significant loss of stability in the FE models. Previous investigations by Keaveny and Bartel (1993) showed a change in loading patterns with loss of collar support. However, the loss of stability seen in the current investigation seems large. Clinical investigation into the effect of collar loss (Manley et al. 1995, Meding et al. 1997) showed no significant decrease in implant stability. The large loss in stability

predicted by the FE may be due to the single load application used. In the clinical situation it is likely that an initial 'bedding in' would occur, such that the relative displacement from the unloaded state would be less than that from the initial position. The bedding in process was not modelled in the FE, so values of micromotion should be treated with caution. It was also found in the prospective clinical study by Meding et al. (1997) that the collarless stems were more prone to distal pedestal formation than the collared version. This seems to agree with evidence found from the FE investigation.

The clinical evidence, for the stem lengths investigated, was limited because such implants have not been used extensively. Where there was evidence it was largely anecdotal, and so of limited use. However, the general trends seen in the FE models seemed to agree with the available evidence. And the mechanical consequences of the design changes imposed seem quite logical.

2.7 Summary of design investigation

The aim of this investigation was to explore the behaviour of the different groups of conservative implants. A broad range of conservative implants were first classified by their overall geometric design. The different groups were then evaluated in a comparable manner using FEA. The results showed patterns of behaviour that were comparable to the available clinical evidence. The investigation showed that different lengths of stem may be prone to different types of failure.

No significant threat was identified in terms of fatigue failure for conservative implants under normal loading conditions. The more sinister type of failure scenarios seemed to be associated with the threat of loosening and bone remodelling. These issues in themselves are also a common source of failure in traditional implants. Also, such biological changes may increase the risk of stem fatigue failure through unforeseen load conditions. One of the main challenges surrounding pre-clinical testing is knowing what the worst case scenario will be for a new type of implant. But by definition, for a novel implant, there is little or no relevant clinical evidence to draw from for such tests. Therefore, subsequent investigations have concentrated on predicting potential failure scenarios which may occur in the future.

3 Predictive behaviour investigations

3.1 Aim

The aim of this investigation was to focus on how implants interacted with the biological host bone and cause it to evolve with time. The objective was to describe the failure processes using numerical techniques, and ultimately predict potential failure mechanisms. This would be an invaluable tool for aiding in the process of pre-clinical testing strategies for novel designs, and could also enable greater understanding of mechanical and biological interactions.

3.1.1 Background

From a mechanical perspective it is known that a fibrous tissue layer can compromise the primary stability of implants (Viceconti et al. 2001). An in vivo study of the effects of stem stiffness in goats by Buma et al. (1997) showed significant differences in the interface reactions around implants of different stiffnesses. So it is evident that mechanical factors can play a role in biological loosening. It is also known that the interface conditions can affect the load transfer and thereby affect bone remodelling (Keaveny and Bartel 1993). Hence it was proposed to deal with both of these issues by exploring both bone remodelling and interface remodelling using numerical simulation techniques. Results then were compared with available clinical data on implants for validation purposes.

3.1.2 Focus

For this investigation it was decided to rationalise the large array of implants under investigation, to just those which could provide the most useful information and/or were of most interest. The primary choice for further investigation from the conservative implants was the resurfacing implant and this was based on the following factors:

- The longer conservative devices have been seen to behave in a manner more similar to standard length stems, and it was evident that the risk of proximal bone loss and distal stress bypass was diminished with the shorter stems. Therefore, for the longer types of conservative device, cantilever bending tests may still be appropriate.
- There is a current trend towards re-exploring the use of resurfacing implants as they are of interest from an industrial perspective. This is because hip resurfacing arthroplasty is recommended for those who are likely to outlive a conventional primary total hip replacement (NICE 2002), but it is acknowledged that less is known about the medium- to long-term safety and reliability of hip resurfacing arthroplasty (NICE 2002).

- Historically there is more experience with the resurfacing type of implant, than with the other relatively unusual conservative devices, so the amount of clinical data available to verify any investigations should be more robust.
- In terms of the expected performance of resurfacing implants, they are historically worse performing than traditional implants. Thus an understanding of the failure mechanism would be of importance for future developments in this area. And by virtue of such typical failures, this may enable a greater understanding of the overall mechanism of failure for all implants.
- Also the investigation has highlighted the possible risk of bone loss in the resurfacing designs, and this was perceived to be of higher risk than evidence seen in metaphyseal designs.

3.1.3 Investigation outline

The investigation was explored using resurfacing implants, as it was recognised that such implants were at a higher risk than most in terms of bone remodelling and loosening. This was to enable a better understanding of the failure mechanics of resurfacing implants, which may then be applicable to all implants.

An alternative design of implant with significant clinical experience was also investigated in view of the age, and commensurate limited anecdotal evidence, available for resurfacing implants. This provided additional support and validation for the simulation techniques which were applied.

The investigation evaluated in more depth the available clinical evidence for the chosen implants. The sequence of events which typically caused loosening was investigated. Then, currently available numerical simulation techniques were discussed and evaluated, for both bone remodelling and interface fibrous tissue formation. Simulations were then carried out on the chosen implants, using the most appropriate techniques combined with FEA.

The behaviour of the simulations was then compared to the available clinical evidence, and evaluated for accuracy. Finally, the simulations were tested on some relatively new implants with a view to aiding in predictions of possible failure modes, and preventative testing strategies that may be appropriate.

3.2 Clinical evidence

There now follows a collection of all available clinical evidence regarding the implants of interest. The implants chosen for investigation were the Wagner resurfacing implant and the Alloclassic stem (Zimmer GmbH). The Wagner is a resurfacing implant which has had a relatively high rate of failure. This type of implant is not currently used extensively, but it has a notable amount of clinical experience. The majority of the evidence was available for the cemented version, so this variant was investigated here. This type of implant historically experienced loosening and biomechanical changes may have played a key role in implant failure, and was therefore worth investigating.

Another implant which was used in this investigation was the Alloclassic primary stem. The Alloclassic is an uncemented Titanium alloy implant. Alloclassic results compare well with the best modern cemented THAs results in the Swedish hip registry (Delaunay and Kapandji 2001). The Alloclassic is a successful and popular implant with a significant amount of long-term clinical information available. This made it suitable for inclusion in this study for validation purposes. Where evidence from similar designs of implant was available, this was also utilised for both the resurfacing and primary types of design.

One of the main difficulties in comparing clinical data with the FE results is the disparate nature of the data types. Clinical data is typically in the form of X-rays. These by their nature are variable in quality and the inferences drawn from the information are highly subjective. The variation between patients is also considerable, so a significant sample size is required to be able to distinguish any significant patterns in behaviour. Typically there is also limited control over variables which could affect the outcome (e.g. patient activity and patient weight). Conversely, the data obtained from FEA are explicitly quantifiable and specific to the case being investigated. For this reason a significant amount of clinical data was analysed and any general patterns of behaviour were identified for comparison with the behaviour of the FE models.

3.2.1 Wagner (stemless resurfacing)

In a study of 11 stable retrieved ICLH femoral resurfacing prostheses by Morberg et al. (2001) there was some evidence of formation of fibrous tissue at the bone interface and evidence of central bony column. Soft tissue was most often found in the distal, peripheral parts of the interface. Inflammatory tissue with macrophages was sometimes observed as well. The specimens were all stable from 9-15 years. Particles were found at the periphery of the femoral neck around the distal part of the prostheses.

It has been suggested that the frequent failures of resurfacing implants may have been caused by the damage caused to the femoral head blood supply by the resurfacing procedure which could be leading to osteonecrosis and subsequent failure. This hypothesis was refuted in a histological study by Campbell et al. (2000) who found no evidence of

osteonecrosis in 25 revised resurfaced femoral heads (cemented THARIES implants and cementless PSR implants). UHMWPE debris related inflammation was cited as the primary cause of failure in these cases.

In a study of 28 revisions of failed Wagner resurfacing prostheses by Strens (1986) (cited by Weinans et al. 1993a), a consistent morphology was noted. Typically a thick fibrous layer was found under the medial and lateral sides of the cup rim. It was found that medial bone resorption was more extensive than lateral (Figure 21). The 28 Wagner implants were revised at an average of 5.5 years post-operatively and were from a series of 84 prostheses. Huiskes et al. (1990) noted that the failure behaviour included bone resorption and formation of fibrous tissue at the cement/bone interface. The failure was thought to start at the cup periphery and subsequently extend towards the apex. The mechanism appeared to start at the medial side. In the central part of the head diffuse decrease in bone density was typically found, but an increase in density was found under the apex of the cup. Huiskes et al. (1990) modelled a series of morphologies using simple 2D models, using the histology as a guide. Results indicated that micromotions could progressively increase after mechanical disruption of the bone implant interface under the lateral and medial side of the cup rim.

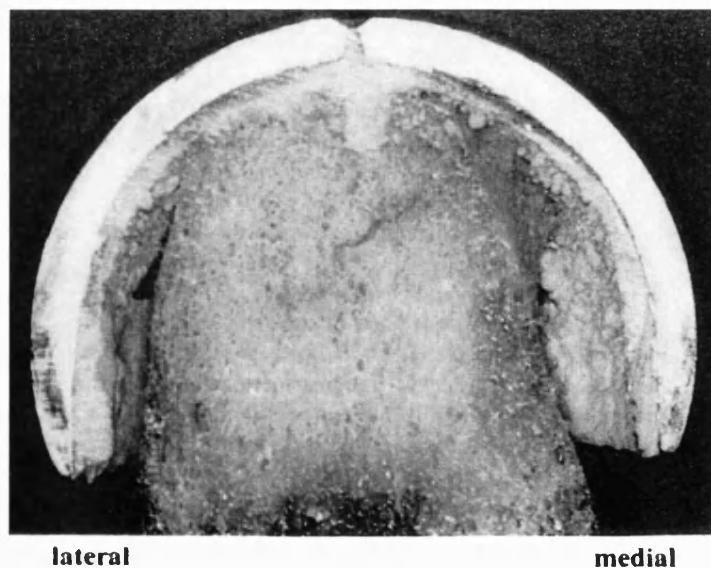


Figure 21: Typical pattern of Wagner behaviour from Weinans et al. (1993a)

Huiskes et al. (1985) indicated that a large amount of bone resorption typically occurred along the periphery of the Wagner implant. The central bony area was usually well attached to the cement layer under the dome of the cup, but occasionally a very thin fibrous layer was present at the interface. The remaining bone was usually sclerotic on the medial side, and osteoporotic on the lateral side. Femoral head collapse sometimes occurred after these extensive morphological changes.

Bell et al. (1985) investigated a series of 18 early aseptic failures of Wagner cemented implants. Loosened femoral components were found to have a membrane at the cement-bone interface. This membrane was thickest at the periphery of the component. Histological examination of the membrane indicated a foreign body response to polyethylene (PE) wear debris. Bone resorption was active at the bone-membrane interface. It was thought that the mechanical failure of the cement-bone interface and the biological response to both particles of cement and fragments of polyethylene seemed to promote loosening.

Ritter and Gie (1986) outlined the clinical results of a series of Indiana resurfacing prostheses. Femoral failures typically exhibited osteonecrosis of the femoral head and femoral neck fracture, although details were limited.

Cserhati et al. (1979) detailed the histology of 3 revision cases of cemented ICLH double cup arthroplasty. The revisions were carried out from 9 weeks to 14 months after initial surgery. A fibrous tissue layer was found to be present between cement and bone. This ranged from 0.5-6mm, and was typically just over 1mm thick, and encapsulated the cancellous bone. Decreased bone density was noted in the specimens. Schreiber and Jacob (1984) also showed detailed histology of these cases. They concluded that bone within the metal shell was probably being subjected to unphysiological loads, resulting in bone resorption and subsequent loosening of the device. However, no investigation was made as to the presence or otherwise of debris related phenomena.

Bradley et al. (1987) histologically examined 25 failed femoral head remnants after ICLH cemented double cup arthroplasty. A fibrous layer was typically present. It was found that osteonecrosis was not inevitable from the use of cemented resurfacing arthroplasty.

Mai et al. (1996) carried out an evaluation of 24 failed femoral THARIES components. Soft tissue was present at the intra-articular margins. The tissue had replaced the bone of the femoral neck, resulting in a reduction in cross sectional area of the neck. Similar tissue was present proximally, which led to variable degrees of cement/bone interface disruption. Evidence of PE and cement debris related inflammatory response and osteolysis was seen.

Willems et al. (1988) investigated 15 femoral head remnants from failed Gerard uncemented double cup prostheses. Histology revealed a smooth collagenous membrane had developed at the interface. Nine of the 15 showed resorption leaving a conical head remnant. The time before revision was 10-50 months for those without resorption and 29-82 months for those with resorption. Thus it was thought that resorption represented a later stage in prosthesis failure. Wear debris was not thought to be the main cause of

resorption in these failures, as little or no inflammatory reaction was observed. It was suggested that resorption was due to mechanical stress shielding.

Nasser et al. (1990) examined specimens of 22 cementless titanium alloy on UHMWPE bearing. Histology revealed evidence of massive debris related osteolysis, from both PE and Titanium.

Amstutz et al. (1994) detailed 5 revised titanium cemented hemi-resurfacing implants at average of 7.8 years. The components were revised due to acetabular cartilage deterioration. Sections showed a thin soft tissue membrane between the bone and cement usually around the neck, and not the dome. Only occasional macrophages were seen, and adjacent bone was healthy. The evidence indicated the periprosthetic tissue reactions occurred in the absence of particulate polyethylene.

Trentani and Vaccarino (1978) detailed results of the Paltrinieri-Trentani resurface arthroplasty, which consisted of a PE cup and a cemented steel femoral component. Failure of the femoral component was due to resorption of the femoral stump concentrically and encapsulation by fibrous tissue.

Howie et al. (1990, 1993) carried out histological evaluations of 72 retrieved Wagner resurfacing prostheses. It was found that interface connective tissue could be present at grossly stable bone/cement interface as well as unstable ones. Thicker regions of connective tissue were apparent around the basal area of the implant, with thinner connective tissue apparent more proximally. PE wear debris was found in the superior region of well fixed components. It was thus suggested that debris can migrate along interfaces, even if well fixed. It was thought that these particles provoked an inflammatory response and subsequent bone resorption.

Claes et al. (1990) examined the histology of 14 revised Wagner resurfacing prostheses at a mean of 5.4 years. Femoral components appeared to suffer from bone remodelling on the medial side which was thought to be caused by stress shielding. In these areas cancellous bone was replaced by fibrous tissue. This eventually led to collapse of the femoral head.

An investigation by Amstutz et al. (1989) detailed 170 revisions of a series of 585 THARIES surface replacements analysis. 87% of the failures were due to aseptic loosening. 64% of these failures were acetabular failures, and 36% of them were primarily femoral failures. Loosened specimens typically exhibited fibrous tissue, medially located membranes, and medial gaps, along with wear debris and macrophages. The use of cement was implicated as the contributing factor in loosening.

3.2.1.1 Discussion of Wagner clinical evidence

It is clear from the studies of the Wagner that both bone resorption and fibrous tissue formation are evident in resurfacing prostheses. Behaviour appears to be similar in nature for stable and loosened prosthesis, with more extreme behaviour evident in the loosened implants. Wear debris related osteolysis is implicated in many cases. A typical pattern involves diffuse bone loss in the head, combined with a central bony column typically with increased density under the apex of the cup. Fibrous tissue appears to initiate at the periphery of the cup and advance towards the apex, which starts at the medial edge and progresses to the lateral side, eventually leading to gross loosening and clinical failure.

3.2.2 Alloclassic (primary stem)

Wick and Lester (2004) compared the clinical behaviour of 79 Alloclassic stems with a matched group of 79 Endoplus stems. The Endoplus stem is almost identical to the Alloclassic, being second and third generations respectively of the Zweymüller designed stem, with minor changes in external form. The follow-up period for both groups was between 2 and 10 years. The radiographic observations of incidences and radiolucencies and bone atrophy for the Alloclassic are shown in Table 5. The Endoplus showed a significantly higher incidence of radiolucent lines and stress shielding than the Alloclassic. Radiolucent lines were not seen until the second post-operative year. Subsidence was found not to correlate with the incidence of radiolucent lines. No patient required revision surgery. The radiolucencies mostly appeared in the proximal femur and did not progress. Radiolucencies did not appear to influence the clinical outcome. Cadaver retrieval showed osseointegration of the Alloclassic despite advanced age, osteoporosis and rheumatoid arthritis.

Gruen Zone	1	2	3	4	5	6	7
<i>Incidence of Radiolucencies</i>	5.9%	0%	0%	0%	0%	0%	7.1%
<i>Incidence of Atrophy</i>	22.6%	2.4%	1.2%	0%	0%	1.2%	28.6%

Table 5: Radiographic evidence for Alloclassic

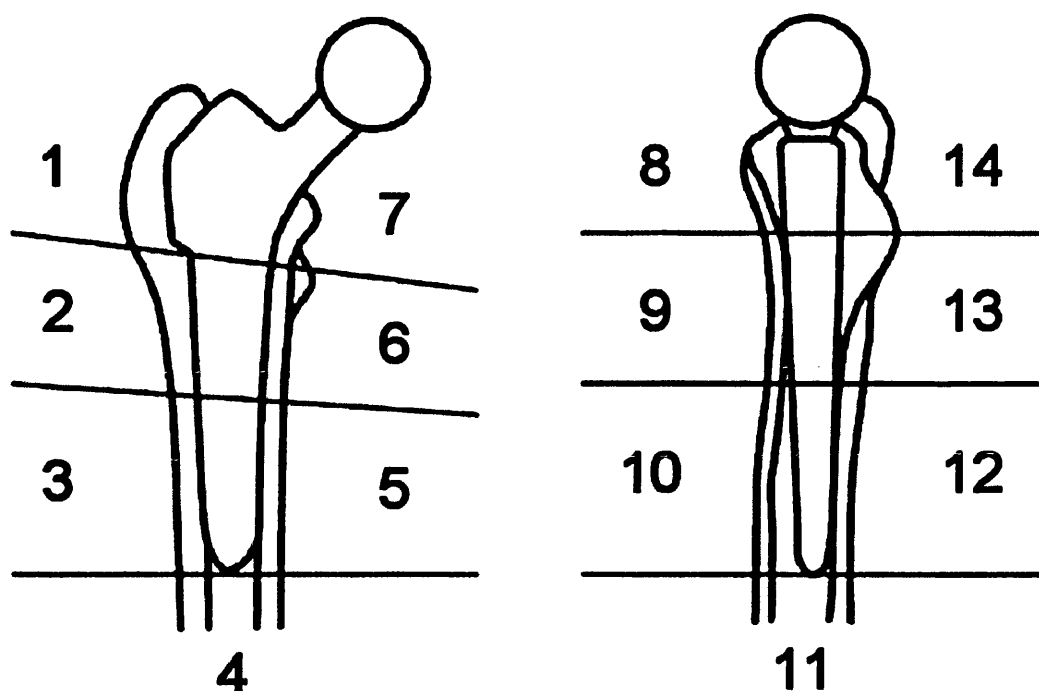


Figure 22: Gruen zone numbering adapted from Pieringer et al. (2003b)

Vervest et al. (2003) carried out a DEXA study on 32 patients implanted with first and second generation Zweymüller designs. The mean follow-up was just over 10 years. The contralateral unimplanted femur was compared with the implanted femur, and this was felt to be a valid comparison. Calcar atrophy was noted in zone 7 in 72% of cases. Hypertrophy was noted in zone 4 in 72% cases. Radiolucent lines were found in Gruen zone 1 in 44% of cases. On average the zone 1 radiolucent lines extended 30mm in length and 2.6mm depth. Radiolucent lines were identified in zone 7 in 31% of cases, extending 32mm on average with an average depth of 1.5mm. It should be noted that this study encompassed the first and second generations of Zweymüller, and no difference was discernible between the two designs, but other authors have found better results with the second generation (Delaunay and Kapandji 2001).

In study by Pieringer et al. (2003a) 74 Alloclassic prostheses were followed-up, with a mean follow-up period of 132 months. The survival rate for the stem was 100%. Radiolucent lines and osteolysis was found proximally, but seldom distally. The observed incidences of osteolysis, radiolucencies and bone atrophy from radiographs are shown in Table 6 and Table 7. Cortical hypertrophy around the stem was seen in 59% of cases. A significant correlation was found between PE liner wear and the incidence of proximal osteolysis. The mechanical distribution of load in the stem was said to lead to bone atrophy in the proximal zones and hypertrophy occurred at the tip of the prosthesis. Proximal radiolucent lines were attributed to micromotions induced by weakened proximal bone stock.

Gruen Zone	1	2	3	4	5	6	7
Incidence of Osteolysis	27%	9%	1%	0%	1%	4%	24%
Incidence of Radiolucencies	45%	5%	3%	1%	3%	4%	42%
Incidence of Atrophy	31%	22%	4%	5%	3%	16%	27%

Table 6: Radiographic evidence for Alloclassic (anterio-posterior)

Gruen zone	8	9	10	11	12	13	14
Incidence of Osteolysis	23%	4%	1%	1%	1%	3%	20%
Incidence of Radiolucencies	43%	5%	1%	3%	1%	5%	36%
Incidence of Atrophy	26%	12%	5%	5%	5%	14%	26%

Table 7: Radiographic evidence for Alloclassic (medio-lateral)

In a radiographic study by Garcia-Cimbrelo et al. (2003) with a mean follow-up duration of 11.3 years, 104 Alloclassic hips were analysed. It was found that 27% had some form of bone pedestal formation at the tip of the prosthesis. Cortical thickening was seen in 30% of hips. Proximal femoral osteopenia was seen in 46% of hips. Femoral osteopenia was found to correlate with the amount of acetabular wear. Mild osteolysis was found proximally in 17% of hips. Subsidence was noted in 12% of hips, but was non-progressive and the stem subsequently became stable.

In a follow-up of 115 Alloclassic by Dohle et al. (2001) after a mean of 8.1 years implantation, the area of the distal stem was characterised by cortical hypertrophy. Radiolucent lines were frequently observed in zones 1 and 7, and this was found not to impair function.

Delaunay et al. (2001) carried out a review of 118 Alloclassic implants with an average of 7.3 years follow-up. Radiolucent lines were always less than 2mm thick and were noted partially in zones 1 and 7 in 36% of hips. These extended completely in zones 1 and 7 in a further 8% of cases, and extended in to zones 2 and 6 in a further 2% of cases. Slight proximal osteopenia was seen in 8% of cases, and a further 3% hips exhibited more extensive loss and showed some evidence of a distal pedestal. Femoral cortical thickening was observed in zones 3, 4 and 5 in 30% of cases, in zones 5, 6 in 10% of cases, and in zones 2 and 3 in 9% of cases. 41% of stems demonstrated no cortical bone remodelling. Calcar atrophy was observed in 77% of cases. There were no revisions for femoral loosening.

Brodner et al. (2004) carried out a longitudinal DEXA study of 100 Alloclassic patients over a 5 year period. DEXA scans were taken at six monthly intervals over the course of

the analysis. The overall changes observed in bone mineral density over the 5 year period are shown in Table 8.

<i>Gruen Zone</i>	<i>1</i>	<i>2</i>	<i>3</i>	<i>4</i>	<i>5</i>	<i>6</i>	<i>7</i>
<i>Change in BMD</i>	-3%	+11%	+1%	+3%	+11%	-6%	-14%

Table 8: DEXA evidence for Alloclassic

Pieringer (Pieringer et al. 2003b) carried out a radiographic study of 47 hips in older patients implanted with the Alloclassic, with a mean follow-up of 69 months. In 23% of stems analysed there was evidence of bone apposition at the tip. The observed incidences of osteolysis, radiolucencies and bone atrophy from the radiographs are shown in Table 9 and Table 10.

<i>Gruen Zone</i>	<i>1</i>	<i>2</i>	<i>3</i>	<i>4</i>	<i>5</i>	<i>6</i>	<i>7</i>
<i>Incidence of Osteolysis</i>	2%	0%	0%	0%	0%	0%	0%
<i>Incidence of Radiolucencies</i>	4%	0%	0%	0%	0%	2%	6%
<i>Incidence of Atrophy</i>	60%	47%	32%	6%	17%	21%	34%

Table 9: Radiographic evidence for Alloclassic (anterio-posterior)

<i>Gruen Zone</i>	<i>8</i>	<i>9</i>	<i>10</i>	<i>11</i>	<i>12</i>	<i>13</i>	<i>14</i>
<i>Incidence of Osteolysis</i>	0%	0%	0%	0%	0%	0%	0%
<i>Incidence of Radiolucencies</i>	4%	0%	0%	0%	0%	0%	4%
<i>Incidence of Atrophy</i>	30%	11%	2%	0%	0%	11%	23%

Table 10: Radiographic evidence for Alloclassic (medio-lateral)

Grubl et al. (2002) analysed 107 hips radiographically, which were implanted with the Alloclassic stem. The average follow-up was approximately 10 years. The observed incidences of radiolucencies and bone atrophy/hypertrophy from the radiographs are shown in Table 11 and Table 12. Osteolysis was not detected around any stem.

<i>Gruen Zone</i>	<i>1</i>	<i>2</i>	<i>3</i>	<i>4</i>	<i>5</i>	<i>6</i>	<i>7</i>
<i>Incidence of Radiolucencies</i>	23%	3%	0%	0%	0%	3%	8%
<i>Incidence of Atrophy</i>	51%	3%	-	-	-	2%	7%
<i>Incidence of Hypertrophy</i>	-	3%	9%	26%	44%	5%	-

Table 11: Radiographic evidence for Alloclassic (anterio-posterior)

<i>Gruen Zone</i>	<i>8</i>	<i>9</i>	<i>10</i>	<i>11</i>	<i>12</i>	<i>13</i>	<i>14</i>
<i>Incidence of Radiolucencies</i>	<i>36%</i>	<i>2%</i>	<i>0%</i>	<i>0%</i>	<i>0%</i>	<i>3%</i>	<i>23%</i>

Table 12: Radiographic evidence for Alloclassic (medio-lateral)

3.2.2.1 Discussion of Alloclassic clinical evidence

The Alloclassic has been shown clinically to be a very successful implant (Delaunay and Kapandji 2001). Typically there was some evidence of radiolucent lines proximally in Gruen zones 1 and 7. Osteolysis has also been found in association with these areas, and has been attributed to PE wear debris. There was also typically evidence of bone atrophy proximally in zones 1 and 7, and evidence of cortical hypertrophy and thickening distally around the stem. There are also some cases which exhibit pedestal formation at the tip of the stem. This suggests that the implant causes stress by-pass proximally, and fixation and stability is primarily achieved through the tapered interlock of the distal stem in the medullary canal. The radiological symptoms exhibited do not appear to have a significant effect on the longevity of the implant, which has very good long-term results.

3.3 Characterising bone

3.3.1 Structure of bone

Bone is a complex biological structure comprising mainly hydroxyapatite mineral, collagen and water. The properties of bone depends on the relative amounts of these substances, and also the architectural organisation of the structure. Bone is also a dynamic structure which can remodel. Remodelling removes old bone and replaces it with new bone. This prevents the accumulation of fatigue damage in the structure, and is thought to optimise the mechanical efficiency of the bone (Martin et al. 1998). Remodelling occurs via Basic Multicellular Units (BMUs) (Frost 1987) which can tunnel through bone and lay down new bone via the action of osteoclasts and osteoblasts respectively.

The porosity of bone can vary continuously, but the two main types are cancellous and cortical bone (Figure 23). Cortical and cancellous bone are made up of the same constituents, but have different structural organisations. Cortical bone has a lower porosity than cancellous. Cancellous bone typically has a high porosity and is composed of struts called trabeculae which are about 200µm in diameter. Remodelling of trabeculae occurs in a similar manner to cortical bone, except that the BMUs can only dig and refill trenches on the trabecular surface (Martin et al. 1998).

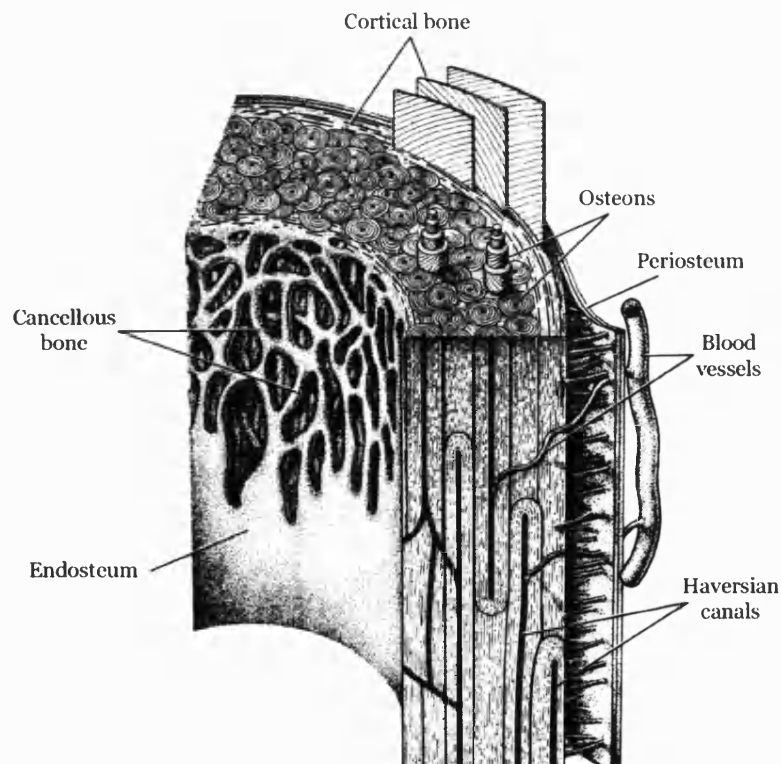


Figure 23: Diagram of typical long bone structure [adapted from Martin et al. (1998)]

3.3.2 Mechanical properties

For both cancellous and cortical bone the properties are anisotropic due to bone architecture. It has been found that bone material properties exhibit orthotropic symmetry associated with the bone architecture (Cowin 2001). Typical properties of human cortical bone are given in Table 13.

<i>Elastic Modulus</i>	<i>Value</i>
<i>Longitudinal</i>	<i>17.4 GPa</i>
<i>Transverse</i>	<i>9.6 GPa</i>
<i>Shear</i>	<i>3.5 GPa</i>

Table 13: Typical mechanical properties of human cortical bone (Cowin 2001)

The mechanical properties of bone vary with porosity (and hence density). Cancellous bone is less stiff than cortical bone. The modulus of bone, E , has been found to depend on the density, ρ , raised to an exponent ($E=a\rho^b$). The value of the exponent, b , has been found to be approximately 2 in various investigations (Figure 24). Regression analyses have shown that the density (or porosity) can account for 70-80% of the variability of the elastic modulus, with other factors such as trabecular orientation contributing to the variation in material properties. Variations also occur between species.

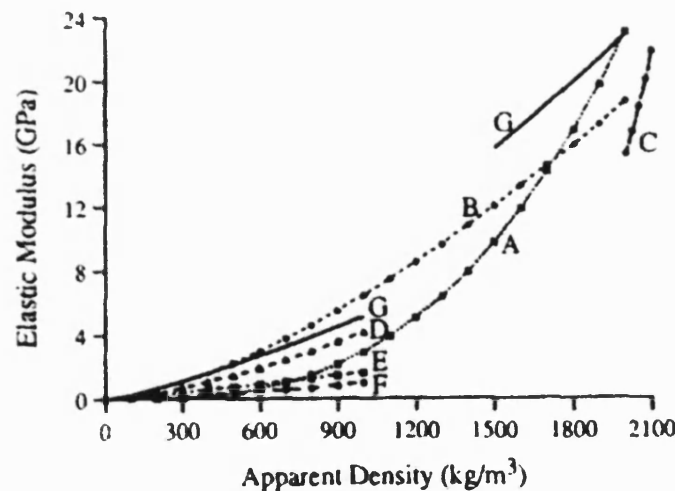


Figure 24: Various relationships (A-G) derived by different authors from testing of human and bovine bone samples [from Rho et al. 1995]

Bone strength has also been found to vary in a similar manner to its modulus (Martin et al. 1998). Although both strength and modulus are dependent on density, it has been found that yield strains are approximately constant over a wide range of densities (Martin et al. 1998). Shear failure strain has been found to be independent of bone density in bovine samples (Ford and Keaveny 1996), with ultimate strains of approximately 4%.

3.4 Bone remodelling theory

3.4.1 Aim

The aim of this section was to evaluate current understanding of bone remodelling theory, and current techniques used to simulate bone remodelling numerically. A suitable technique was then chosen to use for the current investigation into prediction of implant failure scenarios.

3.4.2 Introduction

Wolff is generally acknowledged as being one of the first to publish observations of the adaptation of bone to its mechanical environment (Wolff 1892). Wolff noted that a relation exists between stress transfer and the architecture in bones, and that the bone is able to adapt its structure accordingly. More recently it has been shown that cancellous bone adapts to mechanical loading by changing stiffness (Morgan et al. 2004), in accordance with Wolff's Law. However, the underlying processes of bone remodelling are still not completely understood.

It has been proposed that bone remodelling occurs via Basic Multicellular Units (BMUs) (Frost 1987) which can tunnel through bone and lay down new bone via the action of osteoclasts and osteoblasts respectively. It has been postulated that these BMU's may be activated by micro-cracks occurring as a result of fatigue micro-damage in the bone (Martin 2002).

3.4.3 Simulation

This complex biological process has been somewhat simplified for application to analytical models. FE bone remodelling simulations are typically based on a strain adaptive remodelling theory, such a that a strain based feedback signal controls the density of the bone in the model. Several different feedback signals have been proposed, but one of the most commonly used is Strain Energy Density (SED) (van Rietbergen et al. 1992, Huiskes et al. 2000). A recent experimental study by Kim et al. (2004) found that levels of SED correlated significantly with levels of bone remodelling. The site specific adaptive elasticity type method has been show to give realistic results for a variety of implants when FE simulations and clinical results have been compared (Turner et al. 2004). A recent study by Taylor et al. (2004) showed that the methods used here, can produce clinically relevant results. The SED signal (s) is typically defined as follows in terms of stress (σ), strain (ϵ), and density (ρ):

$$s = \frac{\sum \frac{1}{2} \sigma \cdot \epsilon}{\rho}$$

Rubin et al. (1999) showed directly in vitro that strain is inhibitory to osteoclast recruitment, and thus prevents bone resorption. It was also found that inhibition of

osteoclast formation is achieved locally and was dose dependent, and was not transmitted by a soluble factor. This appears to confirm the viability of site specific strain based formulations of bone remodelling simulations.

Damage based remodelling analyses have also been used to simulate bone remodelling processes. It has been shown by McNamara et al. (1997) that the damage based approach can be equivalent to the strain based approach in remodelling simulations. With strain adaptive remodelling, empirical relationships account for the non-linear behaviour whereas using a damage based approach the non-linearity can be derived theoretically.

FE methods have typically relied on a continuum description of the bone, as the actual trabecular structure is highly complex. Recently, some high resolution small scale representations of the trabecular architecture have been developed which have the ability to control density and also trabecular alignment (Huiskes et al. 2000).

The main driving force behind the remodelling is linked to the loading on the bone (Cowin 1989) or some cyclic variation thereof. Thus FE approaches have generally concentrated on one load case – the peak loading during the gait cycle (Duda et al. 1997). Most investigations use static analysis of the peak loading as a representation of the dynamic system (van Rietbergen et al. 1992). Others have used an averaged daily stress stimulus to give a continuum level stress to account for the different loads applied (Beaupré et al. 1990).

The difficulty with the vast majority of the bone remodelling simulations developed is that of validation. Due to the plethora of variables which can affect the clinical outcome of any particular device, direct comparison with clinical data is difficult. Not least of these variations is caused by individual patient bone morphology, which can vary widely. Advances in CT scanning, digital image analysis and 3D modelling and meshing have enabled the construction of patient specific bone models. The combination of a patient specific model with bone remodelling simulations has recently been compared with the clinical followup after the implantation of a novel THR component (Schmitz et al. 2004). This investigation showed FE simulations using an empirical SED method could produce comparable patterns of bone remodelling to the clinical situation, although it was noted that the confidence in the quality of the clinical data was limited.

3.4.4 Choice of method

For the subsequent investigations, the bone remodelling simulation was based on the methods used in previous bone remodelling investigations, which have been shown to be able to produce realistic representations of patterns of clinical behaviour (Taylor et al. 2004, Schmitz et al. 2004), and can be advantageously implemented in a standard finite element model.

3.5 Pathogenesis at the disrupted bone interface

THR failure caused by aseptic loosening is a common occurrence. The sequence of events which occur to produce loosening at the bone/implant interface is not well defined. Examinations of the failed prostheses have commonly described the presence of fibrous tissue and wear debris at the bone interface. There has been significant discussion over the relative roles of biological and mechanical factors in the disruption of the bone/implant interface. Some investigators have maintained that the cause of such loosening was mechanical in nature whilst others have postulated that biological reactions to foreign materials can lead to failure. There is also an association between the formation of fibrous membranes and the occurrences of periprosthetic osteolysis. There follows a discussion of the available evidence on the development of fibrous tissue at the disrupted bone interface, to clarify the sequence of events which cause loosening.

3.5.1 Evidence about biological initiation

In a histological examination of failed acetabular components, Schmalzried et al. (1992b) found no evidence for a mechanical mechanism for loosening/fibrous tissue formation. Instead, the resorption and fibrous tissue interposition was thought to be biologically initiated by wear debris, which started at the periphery of the cup and advanced superiorly. It was thought from the evidence that bone resorption occurred as a result of macrophage inflammatory response to small particles of high density polyethylene.

The appearance of radiolucent lines has been associated with an increased risk of aseptic loosening in the long term (Kobayashi et al. 1997, Khalily and Whiteside 1998). In a study by Schmalzried et al. (1993) radiolucencies were noted around all well fixed femoral and acetabular components. Whereas changes induced by femoral components were thought to be due to stress changes, particle induced bone remodelling was cited as the cause for acetabular radiolucencies. The most proximal femoral radiolucent lines were found to be associated with interposed soft tissue associated with debris laden macrophage activity. It was also found that radiolucent lines were not necessarily symptomatic of fibrous tissue, but can instead be indicative of endosteal remodelling, which does not necessarily cause component loosening (Schmalzried et al. 1993).

Maloney et al. (2002) examined long term retrievals of cemented THR components. On the acetabular side radiolucent lines corresponded to fibrous tissue interposition whereas on the femoral side they corresponded to areas of endosteal remodelling. The acetabular components typically had progressive disruption to the bone-cement interface, with fibrous tissue interposed. In the femoral components the most proximal 1-2cm of interface was found to have fibrous tissue, whereas the rest of the implant was normally osseointegrated. It was noted that 2cm of fibrous tissue occurring in the acetabular component led to significant destabilisation of the component, whereas the same amount

in the femoral component did not have a significant effect on implant stability. It was acknowledged that the differences between acetabular and femoral loosening were in part due to geometric and loading environments, which would in turn influence access of particulate debris to the implant-bone interface. This evidence suggested that wear debris had a major influence on the breakdown of the interface, but that geometric/loading factors also play an important role.

Howie et al. (1988) carried out an *in vivo* study on rats to investigate the effects of PE wear debris. A fibrous tissue membrane was induced at the interface between bone and cement, using repeated injections of high density polyethylene particles, in an unloaded environment. From the results it was concluded that wear debris in the absence of mechanical load could cause fibrous tissue interposition.

Jasty et al. (1990, 1991) studied retrieved cemented femoral components. Early mechanical failures of the cement-metal interface were found, as were fractures of the cement mantle from cumulative damage. Fibrous tissue was found in association with macrophages, giant cells and particulate fragments of implant material. From these observations it was hypothesised that fibrous tissue at bone/cement interface was secondary to loosening at the implant cement interface. It was thought that local cement fractures caused fragmentation of the PMMA which could lead to local destruction of the cement/bone interface due to focal osteolysis.

In retrieval studies of the interface tissue from 18 aseptically loose femoral components (Horowitz and Purdon 1995, Horowitz et al. 1993) it was found that the process of cellular recruitment in aseptic loosening was caused by PMMA particles. The particles were found to be engulfed by macrophages leading to the production of pro-osteolytic factors leading to bone resorption.

Jasty et al. (1986) analysed osteolytic lesions in well fixed cemented components. Polyethylene wear debris was not found, but PMMA debris was found in association with macrophages and giant cells. The production of PMMA debris was attributed to either micromotions or microfractures due to fatigue loading, as there was no sign of gross cement failure in these cases. There was no sign of a fibrous tissue layer in association with the osteolytic regions around these stable implants.

A more recent study by Shardlow et al. (2003) showed that particles of PMMA generated at the stem/cement interface caused production of pro-osteolytic chemicals which were implicated in the pathogenesis of periprosthetic osteolysis when added to human cell cultures *in vitro*. This behaviour was intensified with the addition of radio-opacifiers to the cement.

In a histological study of 34 THA's with bone loss (both osteolytic and diffuse) by Schmalzried et al. (1992a) wear debris was found in association with macrophage activity in all areas of resorption. It was also discovered that joint fluid can penetrate extensively throughout the component interfaces, even in well fixed and ingrown components. Thus it was concluded that the extent of bone loss would depend on the patterns of joint fluid flow around the implant, and the associated dose of transported wear debris.

Three cemented femoral components were retrieved from autopsy and examined by Massin et al. (2004). Components had been implanted for 8/9 years and all were found to be not grossly loose. Histology was performed at multiple sections along the bone. Wear debris was found at all interfaces (bone/cement and cement/implant) and also in the cancellous bone between trabeculae. Particles were also found in the Haversian canals of cortical bone, and in the medullary canal beneath the step tip. This supports the evidence that wear debris can migrate across the cement-bone interface of stable implants, and can also progress through the open porosity of bone itself.

In a study by Vidovszky et al. (1998) on the differences between osteolytic and non-osteolytic membranes, it was found that osteolytic membranes were suffused with abundant UHMWPE wear debris and inflammatory cells. Typically macrophages and giant cells were found with associated destructive enzyme activity. Non-osteolytic membranes were found to have very small amounts of wear debris, without macrophage activity. It was noted that a non-osteolytic fibrous membrane can occur where the implant is loose where limited wear debris is present. It was suggested from the findings that a non-inflammatory membrane forms initially, which is then transformed into an inflammatory membrane upon the addition of particulate debris. Conversely Dowd et al. (1995) showed that gross mechanical instability without particles produces a similarly large increase of macrophage numbers in the membrane. Dowd et al. also found that the addition of particles to a stable prosthesis also caused an increase in macrophage numbers. The membranes caused by motion were similar in histological appearance to those caused by particulates. Both types of cellular reaction were similar to those seen in failed THA's. It was observed by Vidovszky et al. (1998) that the difference in results was probably due to the differences in degree of implant motion. Thus it was thought that without large amounts of particulate debris, minimal motion does not tend to lead to an inflammatory membrane.

In a histological study of retrieved acetabular fibrous tissue by Lerouge et al. (1997) the aseptic loosening of ceramic cups was found to be not due to a response to debris generated at the articular interface, but to mechanical factors which lead to the fragmentation of the cement and its associated cement opacifying constituents, which in turn caused an inflammatory reaction.

3.5.2 Evidence about mechanical initiation

Significantly different behaviour from Maloney et al. (2002) was reported for retrieved acetabular components by Bauer et al. (1993). In the study uncemented acetabular cups showed evidence of bone remodelling in response to loading. Typically, bone apposition was found around the rim of the component with some evidence of hypertrophy, but with little bone growth over the dome. A thin lucent line was evident over the dome of the cup, which was shown to be occupied by fibrous tissue. This fibrous tissue was found in the absence of polyethylene particles, and without giant cells, although in some cases rare particles of metal debris were found. This study showed markedly different patterns of acetabular behaviour to those of Maloney et al. (2002) and Schmalzried et al. (1992b, 1993). The evidence of fibrous tissue formation in a different pattern, in the absence of wear debris related reactions, and the associated bone remodelling, suggests that mechanical events may be a source of fibrous tissue formation.

In an *in vivo* study by Radin et al. (1982) a group of 28 sheep were implanted with a cemented THR. The sheep were exercised 5 days per week and sacrificed periodically up to a year after surgery, which allowed temporal sequencing of the loosening events. On serial radiographs no changes were seen until 3 months after operation. The resorption of the calcar femorale became evident and progressed with time, where the maximum loss was 6mm after 12 months. A radiolucent line became evident at three months and progressed at 9 months to complete encapsulation at 12 months. Histologically, by 3 weeks there was a loose fibroblastic layer at the bone surface, which became more fibrous at 6 weeks. At 3 months bone remodelling was noted adjacent to the fibrous layer. At six months the fibrous layer was mildly inflamed. The fibrous layer persisted at 9 months. Bone remodelling was apparent throughout, and cement cracking started to occur at 9 months. Cement monomer toxicity and thermal necrosis was not implicated in loosening of the prosthesis. The evidence suggests that the fatigue failure of PMMA was a secondary phenomenon. Calcar resorption and the formation of a radiolucent line at the bone/cement interface was associated with a decrease in torsional rigidity and histological deterioration of the bone/cement interface. It was noted that previous studies, where the radiolucent line which frequently appears was thought not of importance, was disputed by this study by its association with the formation of fibrous tissue and reduced torsional rigidity of the structure. It was thought from this investigation that either mechanical or vascular effects were the root cause of the loosening of the prosthesis.

Boss et al. (1994) presented a detailed description of the development of the non-inflammatory interface membrane and the inflammatory interface membrane associated with osteolysis, based on examination of previous investigations. It was noted that the non-inflammatory interface membrane was often present between implant and bone and could be up to 1.5mm thick, and it was thought that this variation may reflect varying

degrees of micro-instability. The non-inflammatory interface membrane lacks inflammatory infiltrates and does not mediate bone remodelling, and is often present even in stable prostheses. Thus it was thought that mechanical factors lead to formation of the non-inflammatory interface membrane. The inflammatory interface membrane, which is typically thicker and more inflamed, was usually associated with loosened prostheses and can cause bone resorption through its additional inflammatory cells, such as macrophages and giant cells which have been associated the phagocytosis of wear debris. Thus it was thought that the inflammatory interface membrane was potentiated by biological reaction to wear debris. It is also noted that the histological and biochemical profiles of interfacial membranes are qualitatively similar, for both loose and stable, and for cementless and cemented prostheses.

In a study by Goodman et al. (1989), who studied the fibrous tissue membrane found at the interface of 31 loose/non-loose prostheses at revision, loose components demonstrated statistically higher PGE₂ levels (which mediates bone resorption) in the fibrous layer than those in the non-loose group. The non-loose specimens were associated with statistically fewer giant cells and histiocytes. And the fibrous layer was usually less well developed in the non-loose specimens. This study implicated interfacial motion as the cause of the fibrous tissue.

Mjoberg (1994) supported the theory of loosening induced wear – as opposed to wear induced loosening. It was argued that the theory of early loosening (caused by mechanical factors) could explain both the instances of rapid early migration, and also to a great extent the epidemiology of clinical failure. It was also argued that the hypothesis of wear particles causing prosthetic loosening by foreign-body reaction, could not be true without additional assumptions being required.

Sew Hoy et al. (1981) carried out a canine in vivo study to assess the behaviour of the cement/bone interface. Twenty-seven dogs were implanted with scaled down replicas of the THARIES resurfacing prostheses. The dogs were sacrificed at intervals from 2 weeks to 46 weeks. X-rays showed evidence of radiolucent lines in all but 4 dogs. The 4 exceptions all had a follow-up of less than 8 weeks. Using histology, at 2 weeks after surgery the majority of the interface was interspersed with a thin amorphous layer. After 2 weeks a fibrous layer was consistently present between cement and bone. This varied from a few cells to 2mm in thickness. The depth of fibrous tissue correlated well with cement penetration depth. Where 1-2mm penetration was achieved the membrane was minimal. Conversely, poor penetration resulted in early formation of a thick interface membrane which increased in depth slightly with time. With the thinner membrane osteoclastic activity was decreased. After 4 weeks, development of the fibrous tissue was observed. There were no clinical failures of components. The organisation of woven bone at the interface into horizontal trabeculae was only found with animals with a long follow-up and

a wide radiolucent line. From the evidence in this study, micromotion was cited as the major cause of interface fibrous tissue formation.

Aspenberg and Herbertsson (1996) carried out an *in vivo* experiment on 68 rats, to compare the effects of particles versus movement on the development of a fibrous membrane. It was found that the presence of high density PE particles at the interface alone did not cause fibrous tissue formation, and did not cause bone resorption. Conversely, sliding movements at the interface caused formation of a fibrous membrane. After 2 weeks of movement local areas of soft tissue were seen. At six weeks only small areas of bone contact were remaining, and the tissue had a mean thickness of 0.1mm. This membrane was statistically thinner than the original soft tissue foci. Cessation of movement caused the membrane to change back to bone. The combination of movement and particles had little compounding effect, but did cause the membrane to persist even upon cessation of movement. It concluded that mechanical stimuli are of primary importance for prosthetic loosening, and particles modulate the later stages of the loosening process. Aspenberg disputes previous work by Howie et al. (1988) as having caused arthritis via the massive doses of particles.

Goodman et al. (1994) carried out bone chamber experiments on rabbits to investigate the effects of micromotion on fibrous tissue formation. Histology of unmoved chambers showed extensive trabecular bone after 3 weeks. The specimens exposed to daily micromotion for 3 weeks showed primarily fibrous tissue. Specimens exposed to 3 weeks micromotion followed by 3 weeks without micromotion showed similar histology to the unmoved specimens.

MacQuarrie et al. (2004) carried out *in vitro* experiments to assess the roles of particulates and cyclic mechanical strain on osteoclasts, which are known to resorb bone (Haynes et al. 2004). In static cultures, wear particles increased osteoclast differentiation. Mechanical strain alone decreased osteoclast differentiation. It was found that the addition of mechanical stimulation to a particle laden system enhanced osteoclast survival.

In a series of *in vitro* experiments by McEvoy et al. (2002) cyclic pressure variations were compared in their ability to activate macrophage cytokine production, which has been shown to be involved in bone resorption by activating osteoclasts. The combination of cyclic pressure and particles created a synergistic elevation on the levels of cytokine production. It was also found from this work found that cyclic pressure alone could induce macrophages to cause bone resorption. It was also found that there were varying levels of response to the different pressures from macrophages from different individuals, and this was thought to be due to differential sensitivity of the cells to pressure, where their optimal sensitivity was at physiological levels. It was concluded from this study that cyclic

pressure and wear particles were equally important in the macrophage activation associated with osteolysis around loosened implants. A pressure of 0.0345MPa produced the greatest effect.

In an in vitro study by Rubin et al. (1997) it was found that increased static pressure caused a decrease in osteoclastic activity. Converse to these results, van der Vis et al. (1998) found in a rabbit study that constant saline fluid pressure over 2 weeks led to massive bone resorption in every specimen. It was acknowledged that the pressure was far below the mechanical stimulation of bone remodelling. It was noted by other authors (Skripitz and Aspenberg 2000) that these experiments used saline, which may have affected the chemical environment. In a later experiment (van der Vis et al. 1998) cyclic pressure was applied in a similar manner, and a lesser amount of bone resorption occurred. In a third experiment (van der Vis et al. 1999) pressure was applied by direct mechanical means to a fibrous membrane. This also caused bone resorption in a similar manner. Aspenberg and van der Vis (1998b) noted that in all three experiments bone resorption/osteolysis was preceded by osteocyte apoptosis (programmed cell death), which can spread deeper into the bone. Thus it was argued that the effects of high pressure seemed to kill osteocytes rather than stimulate osteoclasts directly. It is thought that this may be mediated by the flow induced by pressure gradients in the fibrous membrane.

Fujishiro et al. (2004) studied in vitro the effects of mechanical strain and particles on macrophage production of PGE₂, which is an important mediator involved in loosening and osteolysis. The combination of wear debris and movement created a synergistic elevation of PGE₂, compared to wear debris or movement alone. This suggested that mechanical environment combined with particle load could contribute significantly to the development of osteolysis, such that a reduction in cyclic mechanical strain or wear debris could increase implant survival.

In an in vitro study, Sampathkumar et al. (2003) examined the behaviour of macrophages, subjected to wear debris and pressure. They tested the hypothesis that pressure may be the initiating factor for the recruitment of macrophages into the periprosthetic tissue. It was noted from previous studies that macrophages phagocytose particles and release inflammatory mediators, but it had been found that selective blocking of these mediators did not stop bone resorption. The study investigated the effect of other cytokines/chemokines which are important in the recruitment of cells to inflammatory sites. The results showed significant variability in the release of factors by each donor. The secretion of macrophage colony stimulating factor and PGE₂ by macrophages was increased dramatically with exposure to pressure or particles, and the effects were synergistic when applied simultaneously. When looking at the chemokines (which are involved in the recruitment of inflammatory cells) pressure was more dominant than particles and no

synergistic effect was seen. The results suggested that pressure was the initiating factor for the recruitment of cells into the periprosthetic tissue.

A study by Skoglund and Aspenberg (2003) investigated the effect in vivo of PMMA particles and cyclic pressure using 59 rats. Pressure was applied by compression of a fibrous membrane at 0.17 Hz. There was a measurable resorption in response to PMMA particles (with radio opacifier), but no additive or synergistic effect from introducing particles to the pressure model, and the effect of pressure was far greater than that of particles. This suggests that the effects of pressure are far more important than that of particles, in the instigation of the osteolytic response. If the fluid flow had been zero then the pressure used would have been 0.6MPa, but video registration showed that the pressure was only a small fraction of this value.

Bechtold et al. (2002) investigated the effects of PE debris and movement in a canine in vivo study. It was found that if the interface was stable, the presence of PE was not sufficient to produce an osteolytic response or a significant membrane. Only when the interface was unstable did a fibrous membrane form, and then only in the presence of PE was this membrane inflamed. Thus from this study it was concluded that particulate PE did not primarily determine the differentiation response; instead the mechanical instability of the interface caused fibrous tissue formation, which could then become inflamed upon addition of particulates to the unstable interface.

Matthews et al. (2001) carried out in vitro tests on macrophages to study the combined effects of cyclic mechanical strain and wear particles on pro-inflammatory cytokines. The experiment demonstrated synergistic effect from particles and movement. It was found that cyclic compression was not as important as the sub-micron PE particles. It was suggested that at low particle load, micromotion may contribute significantly to macrophage activation and osteolysis in vivo.

Jones et al. (2001) investigated the effect of PMMA particles and movement in an in vivo study on 66 canines. It was found that the initiation of inflammatory response was much slower to particles alone than to mechanical instability alone. Animals exposed to both PMMA particles and movement had an intense inflammatory response associated with accelerated bone loss.

In a histological and immunohistochemical evaluation of interface membranes from loose cemented and uncemented acetabular components (Jones et al. 1999), it was found that cemented and uncemented components suffer from the same loosening pathology. It was noted that although certain cell types and cytokines were found in association with the periprosthetic membrane of loosened implants, their precise roles and interactions in the pathogenesis of loosening was not well understood.

Goldring et al. (1983) gave a detailed description of non-inflammatory fibrous membranes found around 20 aseptic loosened cemented implants at revision. The fibrous membrane was found to be 0.1-1mm thick. It was thought that formation of the membrane via movement was developmentally appropriate given its morphology. It was also found that the membrane had the capacity to mediate resorption of bone via release of PGE₂.

Goodman et al. (1995) investigated the effects of micromotion and particles on 9 rabbits using a bone chamber model, over a 3 week period. It was found that micromotion inhibited bone growth and led to the formation of fibrous tissue. PE particles were associated with macrophages. The combination of both effects led to a fibrous layer laden with macrophages. It was noted that in this experiment, no additive/synergistic effects were found in the short experiment length investigated.

Skripitz and Aspenberg (2000) investigated the effect of pressure induced by loading of the fibrous tissue layer adjacent to bone, with in vitro experiments on rats. Cyclic mechanical pressure of 0.6 MPa at 0.17Hz was applied to a layer of fibrous tissue generated in the bones of 16 rats for a period of 28 days. This caused osteolysis in the specimens. The formation of a walled off cystic lesion was typically observed.

In a similar experiment by Tagil and Aspenberg (1999), a bone chamber which was estimated to produce a compressive hydrostatic stress of 2 MPa at 0.17Hz caused cartilage (either hyaline or fibrous) to form, but no osteolysis. It was noted in this study that loading parameters in the experiment may be critical to the tissue differentiation observed.

A sensitivity to loading parameters was seen by Aspenberg et al. (1996) in a bone chamber study on rabbits. In this experiment the effect of the frequency of load application was investigated. The 1Hz loading tended to increase the amount of ingrown bone compared with no motion. The 0.17Hz loading caused a marked decrease in bone formation and an increase in fibrous tissue formation, without signs of inflammation. It was concluded from this experiment that the formation of fibrous tissue was associated with the parameters of tissue deformation rather than by the amount of tissue damage.

In a study by de Rooij et al. (2001) the behaviour of mechanically induced cartilage was investigated using bone chambers which loaded the tissue in the chamber, with a 2 MPa daily application of cyclic force. In this experiment most specimens showed focal islands of cartilage in a fibrous tissue layer. Bone resorption was present under these islands. It was noted from a retrieval study of tibial prostheses (Giori et al. 1995) that fibrocartilage was found in the areas which were mainly under compression, and fibrous tissue was found in the areas which underwent shearing stresses. Some cases showed a contiguous layer of cartilage and no signs of bone resorption. Thus it was speculated that cartilage was formed to protect the underlying bone from resorption. Cartilage may therefore have a

protective function against the hydrostatic stress which can induce resorption. It was indicated upon comparison with other experiments that hydrostatic pressure initiates bone osteolysis and also chondrogenesis at approximately the same level of stress.

3.5.3 Discussion on the causes of implant loosening

It is clear from the evidence above that both mechanical and biological factors can play a significant role in the disruption of the bone/implant interface and the development of fibrous interface tissue and osteolysis. There has been considerable discussion over the role of mechanical factors in the initiation of implant loosening (Mjoberg 1994). Several authors have shown that loosening appears to be due to mechanical factors or at least dominated by them initially (Aspenberg and Herbertsson 1996, Jones et al. 2001, Sampathkumar et al. 2003, Skoglund and Aspenberg 2003). There is also other evidence which supports the role of wear debris as the dominant factor in loosening (Horowitz et al. 1993, Schmalzried et al. 1992a, Howie et al. 1988). In early studies, the pathogenesis of loosening was not clearly elucidated, because typically the membranes harvested at revision surgery were limited to the later stages of loosening. The apparently contradictory arguments of biological and mechanical origins of loosening are somewhat resolved by more recent in vivo studies, such as that of Radin et al. (1982), Sew Hoy et al. (1981) and Aspenberg and Herbertsson (1996). The relation between the biological and mechanical effects has been further elucidated by controlled in vitro studies such as that of McEvoy et al. (2002) and Matthews et al. (2001). Even with this understanding, the cascade of biochemical reactions and the exact stimulus to which cells respond is still not well understood (Jones et al. 1999). It is apparent that the same mechanism occurs in both cemented and uncemented implants (Jones et al. 1999).

From available information, the pathogenesis of loosening is understood to progress in the following manner:

Evidence suggests that development of fibrous tissue is the primary stage of loosening. It is formed at the bone/implant interface. Its formation appears to be triggered by mechanical stimulation caused by shear between the bone and implant. The method of formation is commensurate with the physical structure of the fibrous membrane, which typically has a fibrous arrangement perpendicular to the implant interface. The fibrous membrane formed in this way is typically non-inflammatory in the first instance. The purpose of this layer appears to be to protect the underlying bone from damaging shear strains, by acting as a mediator between implant and bone. The thickness of the fibrous tissue is typically between 0.1mm and 2mm. The thickness of the layer appears to depend on the local mechanical environment. Reduction in the level of mechanical strain facilitates conversion of the fibrous membrane back into bone.

The non-inflamed fibrous membrane can then develop into an inflamed fibrous membrane. This involves the recruitment and activation of macrophages and osteoclasts to the area. It appears that transition to the inflamed state can be caused by foreign body reaction to wear particles including PMMA, PE, and metal. It is apparent that wear debris can suffuse the bone/implant interface, even in well fixed devices. Macrophages are associated with the phagocytosis of the wear particles and subsequently cause a cascade of chemical reactions. This leads to a thickening of the fibrous membrane and can also lead to resorption of surrounding bone, which is typically seen as areas of osteolysis in grossly loosened implants. It has also been found that increased mechanical stimulation of the fibrous membrane itself can cause osteolysis, via pressure induced apoptosis of osteocytes in the surrounding bone. It has also been found that the effects of pressure and particles can compound the osteolytic effect. The permeability of fibrous tissue is typically around 40 times lower than that of bone (Lewis et al. 1984, Lacroix and Prendergast 2002). Therefore hydrostatic pressure is more easily generated in fibrous tissue than bone. Since hydrostatic pressure is thought to be a cause of pressure induced osteolysis, this is in line with the fact that formation of a fibrous layer typically precedes osteolysis. Recent studies have shown that cartilage can be formed under the same conditions as pressure induced osteolysis, in an attempt to protect the underlying bone from pressure induced osteolysis, and that ancillary variables may dramatically affect whether this occurs. Developmentally, cartilage/osteolysis appears to be a secondary phenomenon which is preceded by the formation of fibrous tissue (Skoglund and Aspenberg 2003, Haynes et al. 2004).

The pathogenesis indicated here was in line with observations that early prosthetic migration was found in those implants which loosened at a later stage (Freeman and Plant-Bordeneuve 1994). This was because wear particles are not present at an early stage, but early prosthetic migration could be associated with the formation of fibrous tissue through early implant instability.

3.5.4 Summary of fibrous tissue pathogenesis investigation

An investigation into the available evidence on the pathogenesis of fibrous tissue was carried out to elucidate the sequence of events which lead to its formation. There has been significant discussion over the relative roles of biological and mechanical factors in the disruption of the bone/implant interface. Both mechanical and biological evidence was evaluated and encompassed in a reasoned description of the loosening process. From the evidence it was found that the biological loosening process and formation of fibrous tissue at the interface were governed by the mechanical environment in a similar manner to traditional bone remodelling theory. It was also noted that the effects of the mechanically induced membrane could be subsequently exacerbated by the effects of wear debris related immunological response, leading to inflammation and osteolysis. Therefore it was deemed worthwhile to simulate the formation of fibrous tissue using mechanical methods.

3.6 Interface differentiation models

There are several hypotheses which have been used to model the development of interface tissue. In order to decide which method would be most suitable for the current investigation, the various methods were evaluated in terms of their clinical relevance, and ease of implementation. All of the methods were similar in that they used shear strain as the stimulus for fibrous tissue, but they varied in their implementation and the tissue stimuli for the formation of cartilage.

3.6.1 Fracture healing method

This method was based on a fracture healing hypotheses as described by Carter et al. (1988), Carter et al. (1998) and Claes and Heigele (1999), whereby the tissue type was determined by the levels of distortional strain and hydrostatic stress in the precursor tissue, similar to that shown in Figure 25. Claes and Heigele (1999) hypothesised that the levels of strain and hydrostatic pressure in the fracture callus determined the differentiation of the callus tissue. FE was used to identify levels of stimuli in the callus structure, and these patterns appeared to be consistent when compared with histological results of an animal experiments.

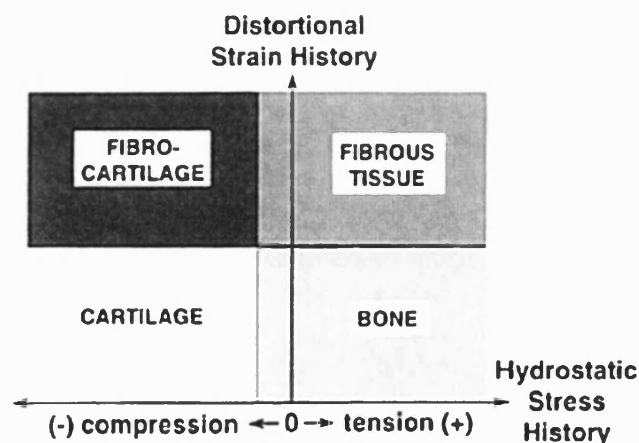


Figure 25: Schematic diagram of fracture healing hypothesis from (Giori et al. 1995)

Giori et al. (1995) investigated the applicability of the fracture healing hypothesis to the formation of tissue at the bone/implant interface. An elastic FE model of a knee was created, and the levels of hydrostatic and distortional strain under a tibial knee plateau were examined. It was found that the locations and levels of the proposed mechanical stimuli thresholds were reflected by the tissue types seen in the clinical situation. It was concluded that a frequently applied distortional strain of 10% was sufficient to stimulate fibrous tissue formation, and that a frequently applied hydrostatic stress of 0.7MPa was sufficient to stimulate cartilage production.

Simmons et al. (2001a, 2001b) investigated the bone/implant interface in more detail and used a unit cell FE method to model the interface behaviour, and compared the results to earlier in vivo experiments of implants in rabbits (Simmons et al. 1999). From the analyses, bony integration was thought to occur when interface tissue strain was less than 8%. The FE analysis indicated that the maximum level of hydrostatic stress experienced was 0.13MPa, and cartilage formation was not observed the experiments. This was thought consistent with a chondrogenesis threshold between 0.15MPa and 2MPa.

3.6.2 Biphasic method

The biphasic theory was proposed by Prendergast et al. (1997), and used both the shear strain and interstitial fluid flow to govern tissue differentiation, as shown in Figure 26. This method was derived from FE models of canine experiments by Søballe et al. (1993) which showed that the shear strain and interstitial fluid velocity varied throughout the tissue differentiation process. It was hypothesised that these variables may regulate the process, rather than being just an effect. Huiskes et al. (1997) developed the proposed regulatory model and detailed a computer simulation which coincided with the tissue differentiation sequence seen experimentally, although it was noted that this did not prove the hypothesis.

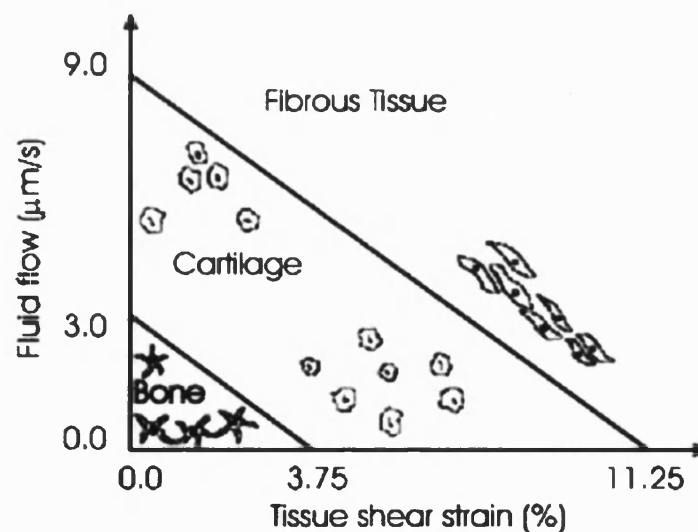


Figure 26: Schematic diagram of biphasic tissue differentiation hypothesis from Geris et al. (2004)

Geris et al. (2004) carried out an FE analysis of a rabbit bone chamber and applied the biphasic theory to create a numerical simulation of the tissue differentiation. The FE results were compared with experimental evidence from rabbit bone chamber experiments. Qualitative agreement of the model was found with experimental findings. It was acknowledged that the stimulus appeared to be dominated by the distortional strain signal.

Yuan et al. (2000) analysed a knee implant model comparable to that analysed by Giori et al. (1995), but with a biphasic analysis instead of a linear elastic analysis. The results supported the hypothesis that interface tissue type was influenced by mechanical variables such as tissue strain and interstitial fluid velocity, but refuted the hypothesis that wear particles were pumped through the interface tissue. The results were similar to those of Giori et al. because of the relatively minute permeability of the tissues.

Lacroix and Prendergast (2002) also used the biphasic model to simulate fracture healing in a callus. The appearance and disappearance of various tissues was similar to histological observation, and reproduced several features of fracture healing.

3.6.3 Shear strain method

Büchler et al. (2003) outlined a theoretical framework by which the development of a model for fibrous tissue at the bone implant interface was carried out. Once the method had been developed, parameters were identified from experimental studies and the method was applied to an idealised model of a bone/hip implant system. The simulation formed advancing fronts of fibrous tissue only at the interface between implant and bone, despite being unrestricted to all of the boney area. The model provided general agreement with clinical observations of fibrous tissue encapsulation (Büchler et al. 2002, 2003), but specific patterns similar to clinical behaviour were not observed. It was thought that this may have been due to the simplified hip model considered. The application of the model to different qualities of bone density, was in agreement with experimental observations on the effect of bone quality (Büchler et al. 2003).

The method assumed that the bone at the interface was initially necrotic, due to surgical trauma, but the bone structure was not significantly modified. Tissue differentiation was then controlled by the shear strain which would govern whether bone or fibrous tissue was formed. Figure 27 shows the tissue differentiation driving function, which determined the tissue type based on the mechanical stimulus, E . The stimulus used for the driving function was the octahedral shear strain. The driving function, $M(E)$, was such that $M(E)=0$ led to bone formation, and $M(E)=1$ led to fibrous tissue formation. The parameters E_{min} and E_{max} corresponded to the thresholds of micromotion 20 μ m and 150 μ m respectively. These parameters were obtained by creating an FE model based on the in vivo canine experimental studies of Jasty et al. (1997) and varying the parameters until similar behaviour to the in vivo experimental data was obtained. This led to values of $E_{min} = 2.3 \times 10^{-3}$ and $E_{max} = 8.2 \times 10^{-3}$. This variation seems reasonable as variable levels of cytokine/enzyme response have been noted for different levels of cyclic pressure in macrophages (McEvoy et al. 2002).

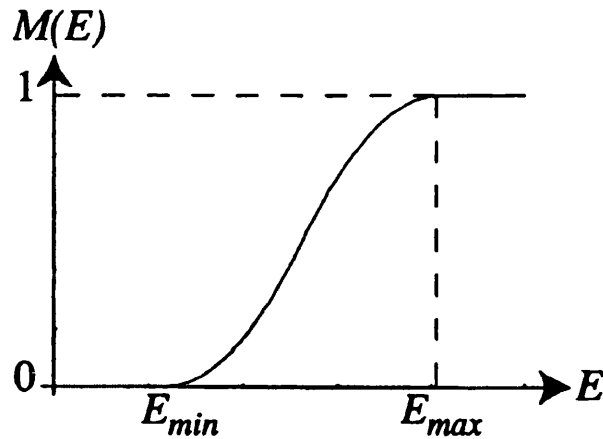


Figure 27: Tissue differentiation driving function (Büchler et al. 2003)

The parameters for the model were based on the evidence of canine micromotion experiments (Jasty et al. 1997, Bragdon et al. 1996, Ramamurti et al. 1997). Jasty et al. (1997) carried out in vivo micromotion experiments on the femurs of twenty dogs. Cylindrical metal implants were press-fitted into the bone and subjected to controlled oscillatory motions, which would theoretically cause zero, 20, 40, and 150 μ m of micromotion at the implant interface over a six week period, and analysed the results histologically after sacrifice. Small movements (zero and 20 μ m) were compatible with stable ingrowth of the bone, whereas the implants subjected to 40 μ m of micromotion were surrounded by patches of fibrocartilage and fibrous tissue. The implants which were subjected to 150 μ m were surrounded by a contiguous layer of fibrous tissue that was one to two millimetres thick.

Bragdon et al. (1996) detailed ostensibly the same in vivo experiments, and reported the change in interface stiffness over the course of the experiment. This was achieved by measuring the torques and resulting displacements dynamically during the application of micromotion. It was found that for the cases with high micromotion, the formation of fibrous tissue was associated with low interface stiffness. Low micromotion values led to an increase in interface stiffness throughout the course of the experiment, which was indicative of bone ingrowth.

Ramamurti et al. (1997) carried out FE modelling of the canine experimental set-up used by Jasty et al. (1997), to investigate the limiting value of micromotion for bone ingrowth. Different values of friction coefficient, press-fit and Young's modulus were investigated, and it was thought from the results that the limiting value of implant motion that inhibits bone ingrowth, could vary with the degree of press-fit.

3.6.4 Discussion of methods

The fracture healing and biphasic methods appear at first to have significantly higher threshold levels of stimuli than the shear strain method. The apparent difference in magnitude of the shear strain thresholds arise from the fact that the elastic and biphasic models are based on the strains developed in the precursor mesenchymal tissue, whereas the shear strain method uses the strain in the bone. More specifically, the shear strain method was said to assume initially necrotic bone at the interface with negligible change in material properties from normal bone (Büchler et al. 2003). Use of the other two methods imply the pre-existence of interface tissue at the interface, whereas the shear strain method does not. Histological studies have shown that an initial phase of bone remodelling can occur at the interface of press-fit implants (Dhert et al. 1998), but it was felt that prescribing the pre-existence of such initial conditions was inadvisable and that that the model itself should govern this behaviour.

Which method gives a more realistic representation of clinical behaviour is obviously of interest. All three hypotheses discussed have shown similarities with animal experiments, on which they are based. Limited qualitative comparisons have also been made for all the methods with interface tissue found in hip and knee implants, and were found to be reasonable representations. The fracture healing and biphasic approaches have additionally been shown to behave reasonably well when simulating the fracture healing process.

A recent study by Geris et al. (2003) compared the results of simulations using the biphasic method and those of the fracture healing method, with animal bone chamber experiment results. From this study it was not possible to identify which method was more accurate in its predictions due to the limited experimental evidence. There has also been some discussion as to which of the elastic or biphasic representations are more substantive (Carter and Beaupré 1999). The biphasic representation seems to give the most well reasoned explanation of how the mechanosensory system may function. But a biphasic model implementation would require significantly increased model complexity as it would necessitate biphasic materials definitions, which have not been explored significantly in these tissues, and also a dynamic force application. It has been shown that the fracture healing approach can give substantially similar results to the biphasic method (Geris et al. 2003).

All three of the methods share commonality in the nature of the parameter which is involved in the formation of fibrous tissue (i.e. distortional shear strain). It is thought reasonable that distortional shear strain is a leading mechano-regulatory factor in the formation of fibrous interface tissue. Goldring et al. (1983) described the appearance of the fibrous tissue structure and considered that formation via movement was developmentally appropriate. It has been noted that the distortional strain component

appears to be more significant than the fluid flow component (Geris et al. 2004), in the biphasic determination of tissue type.

It has been shown that the formation of cartilage is little understood, and sensitive to unexplored variables (de Rooij et al. 2001, Tagil and Aspenberg 1999). The inclusion of cartilage formation could cause increased computational complexity due to its near incompressibility. In several of the animal interface tissue experiments, cartilage formation was not observed. It should also be noted that the formation of fibrous tissue is primary, and typically precedes cartilage formation (see pathogenesis description above). The material properties of cartilage are not significantly different to those of fibrous tissue, especially when compared to those of bone, metal and cement. So any additional global effect of including cartilage formation is likely to be of a lower order than that of fibrous tissue alone.

In view of all these factors it is felt best to dismiss the biphasic and fracture healing approaches in favour of the shear strain method proposed by Büchler et al. (2003). The complex effects of cartilage/osteolysis inducing variables can be neglected, and the primary loosening mechanism of fibrous tissue, can be focussed upon.

3.6.5 Summary

This section has investigated the different hypotheses which have been used to model the development of interface tissue, to decide which method was most suitable for the current investigation. The chosen method was that proposed by Büchler et al. (2003) which uses shear strain as the driving signal which initiates fibrous tissue formation, without considering the complex behaviour of cartilage formation.

3.7 Interface model investigation

3.7.1 Investigation outline

The aim of this investigation was to check that the method proposed by Büchler et al. (2003) performed as expected by testing it on the same set-up as that used to develop it. This involved simulation of the canine micromotion experiments using FEA.

3.7.2 Signal definition

Using the definition for octahedral shear strain proposed by Büchler et al. (2003) and tensor theory, the derivation of the octahedral shear strain was carried out as follows:

For infinitesimal strains,

$$e_{xx} = \frac{\partial u}{\partial x}, \quad e_{yy} = \frac{\partial v}{\partial y}, \quad e_{zz} = \frac{\partial w}{\partial z}$$

$$e_{xy} = \left\{ \frac{\partial v}{\partial x} + \frac{\partial u}{\partial y} \right\}, \quad e_{yz} = \left\{ \frac{\partial w}{\partial y} + \frac{\partial v}{\partial z} \right\}, \quad e_{zx} = \left\{ \frac{\partial u}{\partial z} + \frac{\partial w}{\partial x} \right\}$$

Infinitesimal strain tensor,

$$E = \begin{pmatrix} e_{xx} & \frac{1}{2}e_{xy} & \frac{1}{2}e_{xz} \\ \frac{1}{2}e_{yx} & e_{yy} & \frac{1}{2}e_{yz} \\ \frac{1}{2}e_{zx} & \frac{1}{2}e_{zy} & e_{zz} \end{pmatrix}$$

By definition (Büchler et al. 2003),

$$\varepsilon = \left(tr(E^2) - \frac{1}{3}(trE)^2 \right)^{\frac{1}{2}}$$

Since,

$$e_{ij} = e_{ji}$$

Then,

$$tr(E^2) = e_{xx}^2 + e_{yy}^2 + e_{zz}^2 + \frac{1}{2}(e_{xy}^2 + e_{yz}^2 + e_{zx}^2)$$

and also

$$(trE)^2 = e_{xx}^2 + e_{yy}^2 + e_{zz}^2 + 2(e_{xx}e_{yy} + e_{yy}e_{zz} + e_{zz}e_{xx})$$

Therefore,

$$\begin{aligned} \varepsilon &= \left(\begin{aligned} &e_{xx}^2 + e_{yy}^2 + e_{zz}^2 + \frac{1}{2}(e_{xy}^2 + e_{yz}^2 + e_{zx}^2) \\ &-\frac{1}{3}(e_{xx}^2 + e_{yy}^2 + e_{zz}^2 + 2(e_{xx}e_{yy} + e_{yy}e_{zz} + e_{zz}e_{xx})) \end{aligned} \right)^{\frac{1}{2}} \\ &= \left(\begin{aligned} &\frac{2}{3}(e_{xx}^2 + e_{yy}^2 + e_{zz}^2) \\ &-\frac{2}{3}(e_{xx}e_{yy} + e_{yy}e_{zz} + e_{zz}e_{xx}) \\ &+\frac{1}{2}(e_{xy}^2 + e_{yz}^2 + e_{zx}^2) \end{aligned} \right)^{\frac{1}{2}} \\ &= \frac{1}{\sqrt{3}} \left((e_{xx} - e_{yy})^2 + (e_{yy} - e_{zz})^2 + (e_{zz} - e_{xx})^2 + \frac{3}{2}(e_{xy}^2 + e_{yz}^2 + e_{zx}^2) \right)^{\frac{1}{2}} \end{aligned}$$

This definition differs slightly from a more commonly used alternative, as detailed in (Ford and Alexander 1977)

$$\gamma_{oct} = \frac{2}{3} \left((e_{xx} - e_{yy})^2 + (e_{yy} - e_{zz})^2 + (e_{zz} - e_{xx})^2 + \frac{3}{2} (e_{xy}^2 + e_{yz}^2 + e_{zx}^2) \right)^{\frac{1}{2}}$$

For the purposes of this work the definition derived from the work of Büchler et al (2003) is referred to unless otherwise stated, for compatibility with the previously stated thresholds. For the numerical analysis, values of octahedral shear strain were calculated using principal values where,

$$\begin{aligned} e_{ii} &\rightarrow e_i \\ e_{ij} &\rightarrow 0 \end{aligned}$$

Thus, for the following analyses the octahedral shear strain was defined as follows, in terms of principal strains,

$$\varepsilon = \frac{1}{\sqrt{3}} \left((e_1 - e_2)^2 + (e_2 - e_3)^2 + (e_3 - e_1)^2 \right)^{\frac{1}{2}}$$

3.7.3 Modelling

The analysis was based on canine in vivo micromotion experiments reported by Ramamurti et al. (1997), Jasty et al. (1997), and Bragdon et al. (1996). A similar geometry was used as in the representation of the experiment by Ramamurti et al. (1997), where the cross sectional geometry of the implant in cancellous bone was modelled. This consisted of a 59mm diameter cylinder, which was 10mm thick. This gave a 26mm radial distance from the bone-prosthesis interface to the outer edge. The implant was a 7.2mm diameter cylinder, which was also 10mm in depth. Modelling was carried out using the ABAQUS/CAE software.

Approximately 38,000 ten noded tetrahedral elements were used to mesh the geometry (Figure 28). C3D10M elements were chosen for this analysis, as they are the type of elements used in analysis of bone constructs and bone remodelling analyses. They capture curved geometry well because they are quadratic and they are suitable for contact analyses. The mesh was refined to 0.5mm element edge length at the interface – to reduce anomalies caused by division of the surface into facets. This level of mesh refinement also enabled the behaviour close to the surface to be captured.

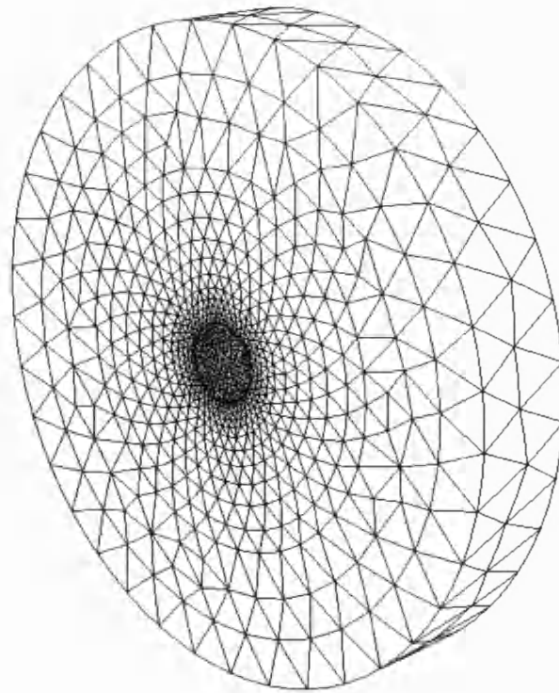


Figure 28: Mesh of canine experiment model

Inferences from canine results

The bone geometry was created with a central hole of 7.0mm diameter whereas the implant had an outside diameter of 7.2mm, as per the experimental geometry. Press-fit conditions were replicated as per the in vivo experiment, giving a 0.2mm diametrical interference fit. Modelling the interference was achieved using a shrink-fit method in the first step of the analysis (ABAQUS/STANDARD). It was noted by Ramamurti et al. (1997) that the results of in vitro experiments comparable to the in vivo set-up indicated that the press-fit did not decay for at least 11 days from the time of implantation, when subjected to micromotion values of 20 μ m. This argued against the possibility of stress relaxation of bone causing a reduction in press-fit. Similar in vitro testing by Bragdon et al. (1996) did reveal that the press-fit could be disrupted by high values of micromotion (150 μ m), probably caused by mechanical damage to the trabecular bone structure. Thus to study the sensitivity to this variable the effect of reduced press-fit for the 150 μ m case was examined.

Jasty et al. (1997) observed the interface between implant and bone microscopically using a calibrated grid, revealing that the bone was in intimate contact with the low micromotion implants, and when the implants were moved, the bone surrounding the implants was deformed and there was no relative motion between the implant and adjacent bone. In view of this evidence the interface was modelled such that no slip would occur.

Jasty et al. (1997) used a microscope and calibrated grid to examine the behaviour of the interface whilst the implants were being oscillated, after the dogs were sacrificed. It was

found that all of the deformation occurred either at the bone implant interface or within the 10 to 20 millimetres of bone surrounding the implants. When the low micromotion cases were examined (zero and 20 μ m), the bone surrounding the implants was deformed and there was no relative motion between the implant and adjacent bone. In the 150 μ m case there was no deformation in the surrounding bone and relative motion occurred between the implant and bone. The 40 μ m case showed patches of both types of behaviour.

The relevance of the no-slip condition to the interface when fibrous tissue was present was uncertain. The evidence from Jasty et al. (1997), as to whether the motion was accommodated by deformation or slip at the interface, was inconclusive. Fibrous tissue would probably exhibit a abraded surface if slip occurred, and this has not been identified explicitly in examinations of fibrous tissue (Goldring et al. 1983). If slip did occur between implant and fibrous tissue, it would be likely that this would have a very low friction coefficient in view of the synovial-like appearance of the membrane (Goldring et al. 1983), and negligible shear would be transferred to the adjacent fibrous tissue. This would mean that the membrane would not be sustainable in any significant thickness, contrary to experimental and clinical evidence. Hence for this model it was assumed that the deformation of the interface tissue accommodated all of the relative motion between the implant and bone (i.e. no slip).

3.7.4 Materials properties

For the purpose of this investigation the materials were assumed isotropic and initially homogeneous. Canine bone materials properties were assigned based on data from Kuhn et al. (1989) from the same bone region as that used in the canine experiments under investigation. Average mechanical properties of canine distal femoral trabecular bone in multiple directions were used (viz. $\frac{1}{3}\{209+158+264\} = 210$ MPa). This lead to an elastic modulus of 210 MPa. Poisson's ratio of 0.3 was used for the bone material, as in a similar investigation by Geris et al. (2003).

The material properties of fibrous tissue were based on those of used by Geris et al. (2003), which were in turn based on experimental measurements by Hori and Lewis (1982). This suggested a Young's modulus of 2MPa which was measured 1 second after loading, due to the highly non-linear response of the material. This timescale was of a similar order to the load duration experienced during walking and was thus felt appropriate. It was noted that the modulus of the fibrous tissue was several orders of magnitude smaller than that of bone or implant materials.

There was limited experimental evidence on the Poisson's ratio of fibrous tissue. In a recent study by Geris et al. (2003) a Poisson's ratio of 0.17 was used, for both the biphasic model, and also the elastic model, although the reason for this was not explicitly justified in the text. This value was found to be typical of a solid cartilaginous matrix when the fluid

was removed (Korhonen et al. 2002). In another study by Giori et al. (1995), the Poisson's ratio for fibrous tissue was inferred from that of cartilage ($\nu=0.42$). The instantaneous Poisson's ratio for cartilage is in the range 0.42-0.49 (Hori and Mockros 1976), which is nearly incompressible instantaneously, due to its low porosity and low fluid flow.

The inconsistency in the Poisson's ratio for fibrous tissue was investigated further and it was found that there was a marked contrast between the properties of cartilage and fibrous tissue. Fibrous tissue takes a long time to re-imbibe any liquid that may be exuded under loading (Hori and Lewis 1982). Under low loads this was seen to take 2-3 minutes, and under high loads this took 10-30 minutes. Cartilage in comparison re-imbibes the liquid much more quickly (Hori and Lewis 1982, Hori and Mockros 1976). It was noted that under the loading conditions under investigation, the time between load cycles was typically a second so the fibrous tissue would not have the opportunity to re-imbibe significant quantities of liquid. Therefore the use of the Poisson's ratio of the solid matrix was felt to be appropriate in this case.

Where a mixture of materials was called for in the interface remodelling algorithm the Voigt 'rule of mixtures' was used (Harris 1999), whereby the properties varied in proportion to the volume fraction of fibrous tissue. The Voigt estimate makes the implicit assumption that the Poisson's ratios of the two materials are equal, which was not the case, but for most practical purposes the difference caused by this error is so small as to be negligible (Harris 1999).

3.7.5 Loading

The outside edge of the bone cylinder was held stationary whilst the implant was rotated by a controlled amount. The displacement was alternately applied and removed to simulate the loaded state and the unloaded state, where the press-fit was re-evaluated. Relative displacements of 20 and 150 μm were simulated. The 150 μm case was then repeated without press-fit as a sensitivity study into the effect of disrupted press fit.

3.7.6 Remodelling algorithm

Variation in fibrous tissue response was prescribed between maximum and minimum threshold values (Figure 27) as was outline previously. The thresholds for formation were as defined by Büchler et al. (2003) (i.e. $E_{min} = 2.3 \times 10^{-3}$ and $E_{max} = 8.2 \times 10^{-3}$). Simulation of the interface tissue differentiation process was achieved with the user subroutine functionality of ABAQUS/STANDARD. The subroutine monitored the level of cyclic octahedral shear strain at each material point of the bone mesh, and altered the material properties locally in response to the level of cyclic octahedral shear strain. Material property transition was controlled such that if the threshold was exceeded then transition to fibrous tissue occurred, and if the signal was below the threshold then transition back to bone occurred. The rate of tissue transition was made proportional to the volume fraction

of fibrous tissue as outlined by Büchler et al. (2003). A comparison of this non-linear transition with a linear equivalent (i.e. constant rate of tissue transition) was carried out, but this revealed no significant difference in the final results. The speed of transition relative to the number of analysis steps was controlled by a fractional multiplier. The remodelling analysis was then repeated in a stepwise fashion for a simulated 3 weeks, or until the simulation had stabilised, as this was the timescale simulated by Büchler et al. (2003). It should be noted that reversibility of the fibrous tissue formation process was accounted for in the model as this had been observed in previous experiments.

3.7.7 Alternative signal definition

Another model was also created during this investigation to simulate fibrous tissue development based on the experimental observations. This method was similar to that of Büchler et al. (2003), but used a different form of the octahedral shear strain signal – the total octahedral shear strain. The difference between the two signal types (cyclic and total) is illustrated in Figure 29.

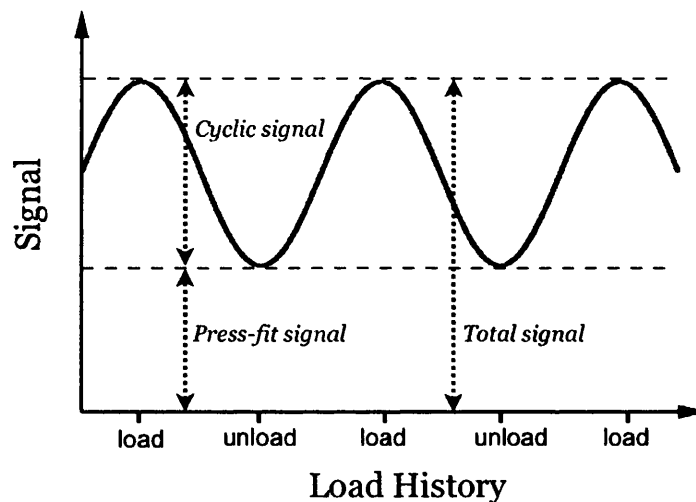


Figure 29: Schematic diagram showing difference between total and cyclic components

Using the alternative, total octahedral shear strain, the press-fit would also be accounted for in the signal. This was felt necessary as there was no previous investigation into the exact nature of the signal and it was not explicitly known whether press-fit should form part of the signal. This method would obviously increase the threshold level for fibrous tissue formation above the level identified by Büchler et al. (2003). Thus for this alternative method the threshold was set using the maximum total octahedral shear strain seen during the 20 μ m case. This single minimum threshold was used as no maximum threshold was easily identifiable, and a step change in the driving function was implemented instead. This was then implemented in a similar manner to the main method.

3.7.8 Main results (cyclic signal)

As expected, the method developed by Büchler et al (2003) did not produce any fibrous tissue at 20µm of micromotion. The remodelling simulation under 150µm of micromotion led to a depth of 2.2mm of fibrous tissue (Figure 30a). The majority of the simulation appeared to stabilised after 3 weeks, although there were still small decreases in tissue depth at this time. The progression of the simulation with time is shown in Figure 30b. The interface stiffness at the end of the simulation was 0.05 Nm/deg.

When the 150µm case without press-fit was analysed this led to similar results, but with a maximum depth of fibrous tissue of 2.9mm. The interface stiffness at the end of the simulation was 0.04 Nm/deg.

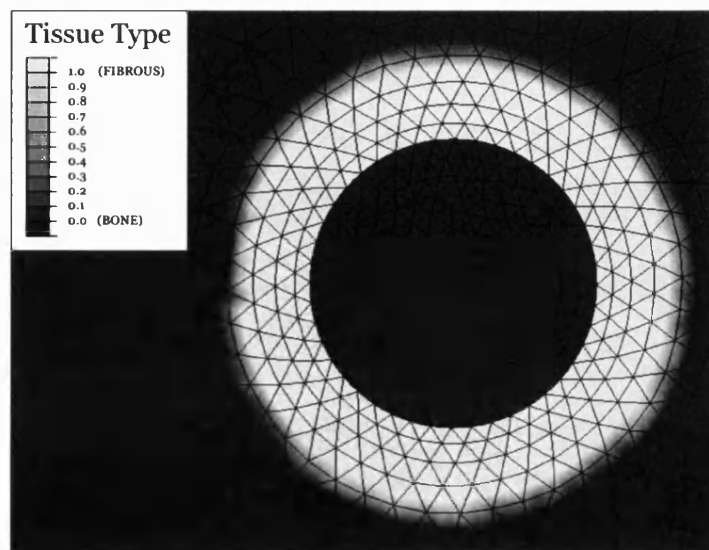


Figure 30a: Fibrous tissue at end of 150µm cyclic simulation (2.2mm)

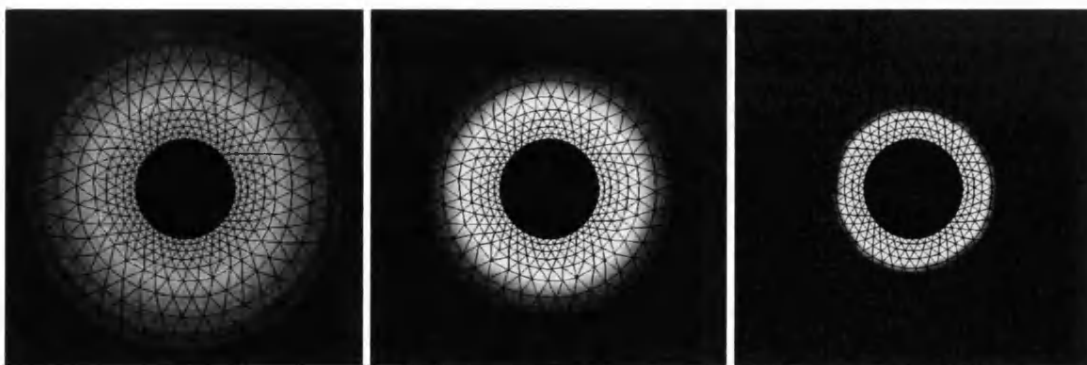


Figure 30b: Progression of cyclic simulation with time (left to right)

3.7.9 Alternative results (total signal)

The maximum octahedral shear strain due to the press-fit alone was 4.4%. This was found at the interface between the implant and bone. The maximum value of total octahedral shear strain under the 20 μ m case was 4.6%. This peak value of 4.6% octahedral shear strain was used as a single threshold for the interface remodelling simulation using the total signal.

The remodelling simulation at 150 μ m developed a contiguous depth of fibrous tissue which stabilised at a maximum depth of approximately 1.2mm (Figure 31a). This layer stabilised after approximately 10 days simulation. The interface stiffness at the end of the simulation was 0.09 Nm/deg.

When the 150 μ m case without press-fit was analysed this led to similar results, but with a maximum depth of fibrous tissue of 0.7mm. The interface stiffness at the end of the simulation was 0.14 Nm/deg.

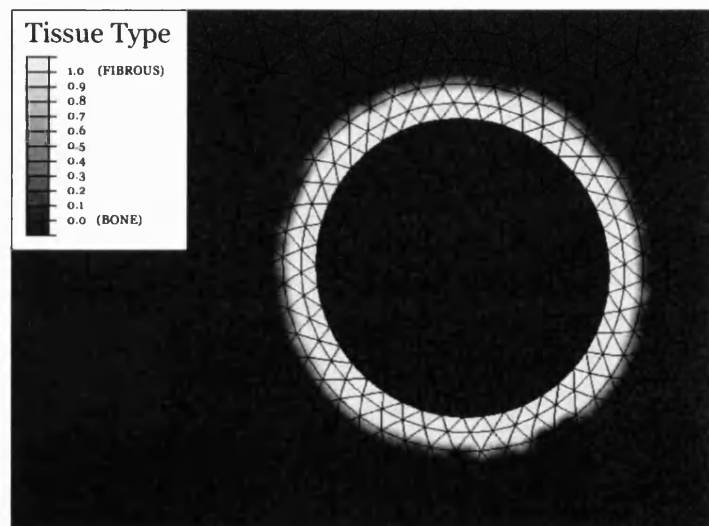


Figure 31a: Fibrous tissue at end of 150 μ m total simulation (1.2mm depth)

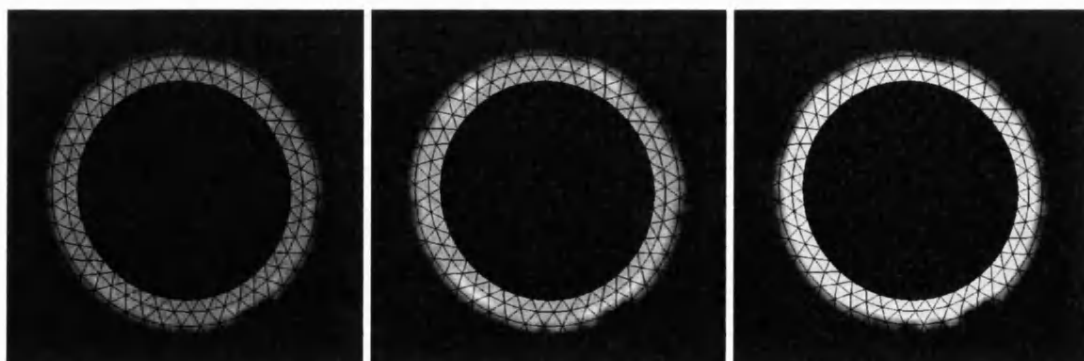


Figure 31b: Progression of total simulation with time (left to right)

3.7.10 Comparison of results with experimental data

The thickness of fibrous tissue observed in the canine experiments ranged from 1 to 2mm in depth according to Jasty et al. (1997), and 0.5 to 1.5mm according to Bragdon et al. (1996). These values are compared to the depths of fibrous tissue found using the fibrous tissue simulation methods in Table 14.

Loadcase	Fibrous Tissue Depth (mm)		
	Cyclic signal results	Total signal results	Canine observations
20µm	0	0	0
150µm (no press-fit)	2.2 (2.9)	1.2 (0.7)	0.5-2.0

Table 14: Comparison of fibrous tissue depths

The interface stiffness was measured at the end of the canine micromotion experiments using a torque sensor (Bragdon et al. 1996). This data was compared with the equivalent stiffness at the end of the FEA simulations in Table 15. The interface stiffness was measured by evaluating reaction forces/moments required to induce the displacements in the FEA at the end of each remodelling analysis.

Loadcase	Interface Stiffness (Nm/deg)		
	Cyclic signal results	Total signal results	Canine measurements
20µm	2.25	2.25	1.25 ± 0.45
150µm (no press-fit)	0.05 (0.04)	0.09 (0.14)	0.16 ± 0.10

Table 15: Comparison of interface stiffnesses

3.7.11 Discussion of results

The simulated 150µm fibrous tissue models (with press-fit) produced depths of fibrous tissue agreed well with experimental observations of fibrous tissue depth from the canine investigations. Progression of the interface tissue development was different between the models based on the cyclic signal alone, and the alternative, total signal. The progression of the cyclic signal driven simulation (Figure 30b) was such that the depth of the initial bone affected was large, and then this decreased in depth to form a contiguous layer of fibrous tissue which was 2.2mm thick, after a simulated 3 weeks. The behaviour of the alternative, total signal driven simulation (Figure 31b) was different, because the fibrous tissue did not vary in depth significantly throughout the simulation, but just changed from bone to fibrous tissue, at a depth of 1.2mm, over a simulated period of 10 days. The pattern of fibrous tissue formation in observed in a different investigation on rats by Aspenberg and Herbertsson (1996) showed a somewhat similar progression to the cyclic

signal driven simulation, where initially thicker foci of soft tissue eventually coalesced into a thinner contiguous membrane.

The effect of reduced press-fit under high levels of micromotion was investigated. This reduced press-fit using the cyclic driven signal caused an increase in tissue depth from 2.2mm to 2.9mm. Conversely, using the total signal driven simulation, this caused a decrease in fibrous tissue depth from 1.2mm to 0.7mm. This change in behaviour under reduced press-fit is reasonable considering the nature of the two driving signals involved. The evidence that press-fit may be lost during high micromotion was compatible with these results, as these results also compared well with the values seen experimentally. It should be borne in mind that the actual level of press-fit achieved during the canine experiment is likely to be less than the design intent, due to variations during the surgical procedure. This may in turn account for some of the variation from the FE results.

Sustainable levels of stress/strain were considered useful, as this may indicate if bone failure was expected to occur under the press-fit conditions. The maximum contact pressure due to the interface shrink-fit was found to be 6.1MPa. The ultimate compressive stress for the bone modelled in the investigation was found to be 6.9MPa using information from Kuhn et al. (1989) (viz. $\sigma_{all} = 0.03 \times E + 0.61 = 6.9\text{MPa}$, where $E = 210\text{MPa}$). This indicated that the press-fit used here generated stresses close to the ultimate compressive stress of the bone in question. Shear failure strain has been found to be isotropic and independent of bone density in bovine samples (Ford and Keaveny 1996) where mean ultimate shear strains were 4.60%. However, similar information was not specifically available for octahedral shear strains, although the values may be expected to be a similar order of magnitude.

Therefore it may be the case that the peak shear strains occurring in this model may be so high as to cause the bone to exceed its yield limits. The post-yield behaviour of bone is currently not very well characterised. The material in this model was considered to be elastic, so behaviour beyond these limits may be unrealistic. However, the overall patterns of where high shear strains are likely to occur are still thought realistic, and therefore a useful indicator of where fibrous tissue may occur. It should also be noted that those regions which may initially go beyond the elastic limits, are likely to undergo a significant reduction in stiffness in the next load step, due to the differentiation into fibrous tissue. Indeed it may in fact be the case that trabecular shear failure may initiate the formation of fibrous tissue. It should be noted that in the canine experiment investigated here, the loading situation was displacement controlled. In the clinical situation the implant environment is typically load controlled, such that increased press-fit would reduce the amount of relative motion between implant and bone and thereby limit the amount of shear strain at the interface.

The interface stiffnesses seen in the models were of a similar order to those measured experimentally (Table 15). Examination of the 150 μ m cases (both with and without press-fit) showed comparable levels of interface stiffness to the experiments. The simulations which produced thicker depths of fibrous tissue, were seen to have a correspondingly lower interface stiffness. The 20 μ m FE measurement (no fibrous tissue) showed an interface stiffness twice that of the experimental reading. This may have been attributable to an initial error in the estimation of the material properties of the bone model. However, rerunning the analysis with a reduced bone modulus did not affect the fibrous tissue simulation results significantly because the analysis was displacement controlled. The modulus which corresponded to the measured interface stiffness was approximately half that of the original modulus estimate. This may be in agreement with the experimental evidence, where it was noted by Jasty et al. (1997) that there was marked atrophy of the trabecular bone surrounding the implants which were subject to 20 μ m or less, relative to the higher micromotion cases, which was thought to be due to the stress shielding caused by the presence of the metal implant. It is understood that the initially uniform isotropic material properties may have affected the results somewhat. Higher shear strains may be expected to be generated in the actual trabecular bone structure due to the local variations in bone density and the anisotropy of the structure.

The levels of total octahedral shear strain appear to be a good indicator for the development of fibrous tissue formation. The exact nature of the signal is not clear from these investigations. In terms of signal generation in the biological tissue, it was felt that both methods could be equally valid. The cyclic octahedral shear signal would be representative of the changes in shape caused by the cyclic component of the signal, such as pulsatile fluid effects. Conversely, the total octahedral shear signal would be more representative of a damage based criterion in the bone structure, where the full deformation including press-fit would be critical.

Both the total and cyclic methods described above give reasonably similar results to the clinical behaviour. The threshold for fibrous tissue formation, using the total signal was calculated to be 4.6%. This is significantly larger than the threshold using the cyclic component alone which was seen to be 0.2-0.8% strain. Whether the cyclic octahedral shear strain, or the total octahedral shear strain is the driving factor is still a matter for debate. The varying progressive behaviour seen between the two methods, under limited press-fit conditions, could provide a useful indicator if a suitable in vitro experiment could be carried out, although there is no scope for this in the current investigation.

To try and distinguish the more appropriate type of octahedral shear strain, the physiological levels of octahedral shear strain were investigated in the human femur. It is known from experimental evidence that the levels of shear strain in the human femur under physiological loading can exceed 0.3% on the surface of the bone (Glisson et al.

2000). Thus the threshold of fibrous tissue formation can be expected to be greater than this, such that typical activities on the unimplanted femur would not cause fibrous tissue formation. Further investigation into this area revealed that the physiological levels of octahedral shear strain found in an FE model of the natural femur under peak gait loading could reach around 1% in the central neck region, with values of approximately 0.5% seen in the greater trochanter (Figure 32). The areas of high shear strain are typically characterised by trabecular arcades in the bone structure. This may be an indication of the the relationship between form and function, as these arcade may be more apt to cope with such shear strains.

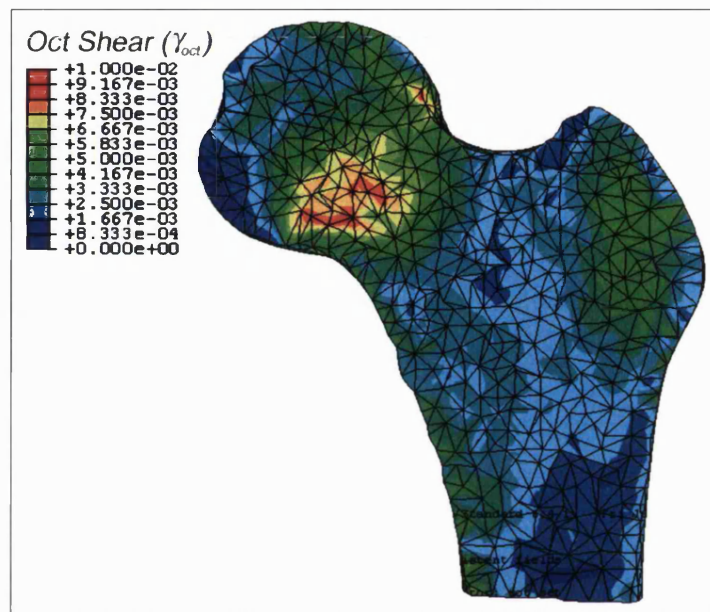


Figure 32: Natural levels of octahedral shear strain in a femoral cross-section

From the above evidence it would at first seem that the threshold level for fibrous tissue formation would be larger than those seen in the physiological situation. Firstly, it should be noted that there may be significant inter-species variation, and that the thresholds seen here are based on canine experiments. Equally, it could also be the case that there is a spatially varying level of sustainable shear strain, over and above which significant increases would cause fibrous tissue formation. This is a similar concept to that used in current bone remodelling simulations, where a threshold (lazy zone), over and above the physiologically normal levels of signal, triggers the biomechanical changes.

3.7.12 Conclusions

From these investigations it appears that the model developed by Büchler et al. (2003) can be implemented in a manner which will produced patterns of fibrous tissue formation similar to those seen in canine experimental studies.

Of the two signal types investigated, there was not enough difference in the behaviour of the models to identify the clinically relevant one. Distinguishing between these using experimental methods was beyond the scope of this investigation. During subsequent investigations the most appropriate method was established by testing both on a different clinical situation to that already investigated and observing which gave the most clinically relevant results.

3.7.13 Summary

The aim of this investigation was to use the chosen fibrous tissue simulation method, on a known experimental set-up to develop and prove correct functionality of the theory and methods. This involved simulation using FE of the canine micromotion experiments, which were used to initially develop the method.

An alternative definition of the simulation driving signal was proposed during the development phase, which was also examined alongside the main method. Both methods provided reasonable simulations of the formation of fibrous tissue, which compared well with experimental evidence. The different methods did vary slightly in the patterns of their behaviour, but there was not enough evidence to discern which was correct. It was decided to use both methods on a different implant loadcase and utilise the method which most resembled the typical clinical behaviour of the case investigated.

3.8 Predictive simulation analysis

3.8.1 Investigation outline

The aim of this investigation was to implement the bone remodelling and fibrous tissue simulations on FE models of the Wagner and Alloclassic implants, and then compare the results with clinical behaviour of these implants. This was then used to judge whether these methods are suitable for use as part of a pre-clinical testing strategy, and to help evaluate their roles in this field.

3.8.2 Method

3.8.2.1 Geometry

The Wagner resurfacing CAD model (Figure 33) was created manually in Unigraphics using design drawings. Implant size was based on the appropriate size for the bone being investigated. Small radii and features (<1mm) which could unduly affect the meshing process were removed. The cemented version of the implant was used, as this was the version for which the majority of clinical evidence was available. A cement depth of 1mm was modelled, without any additional cement penetration into the bone, commensurate with the thin layers of cement seen in clinical evidence. Implant tooling solids were created to carry out the virtual operation. The implant was placed coaxial with the femoral neck, as was commonly seen with this and other similar types of implant, with depth of insertion commensurate with required leg length.



Figure 33: Wagner resurfacing model

An alternative placement was also investigated which followed historical guidance on the placement procedure (Wagner 1978), which advised that the neck angle should ideally be 145° (i.e. valgus component positioning). This required additional geometry to be created with an increased size of device and associated tooling for the virtual operation. It is worth

noting that using this increased neck angle, it was found extremely difficult to carry out the virtual operation without notching the femoral neck, and subsequent cementing was not ideal. This is perhaps symptomatic of some of the early problems encountered during surgery with this type of implant, and subsequent neck fractures which have occurred (Bradley et al. 1987). The operational technique and also the method itself have been implicated in these neck fractures (Spranger and Eder 1980).

An appropriately sized Alloclassic CAD model was used based on the size of the bone. Custom implant tooling solids were created to aid in the virtual implantation into the bone model. The implant was placed in the centre of the transected neck of the femur with the stem pointing down the medullary canal (Figure 34). The depth of insertion was commensurate with required leg length. A typical 28mm ball head was included in the model to take account of any effects caused by the offset in loading. It was found that the positioning of the component in this case would have involved transection of the Piriformis and Gemelli muscle groups, thus these muscles were not included in the loading of the implanted model.



Figure 34: Alloclassic model

The Alloclassic implant is typically implanted in the femur with a certain level of press-fit during surgery. It was unknown whether this press-fit persists in the long-term. There is

some evidence that the pre-loading applied in other implants during surgery can subside (Bereiter et al. 1991), due to the viscoelastic response of bone. However, the Alloclassic's design is such that self tightening may occur, making relaxation less likely. The effect of press-fit may have on the simulations is unknown, and it may be significant. This may be especially important for the fibrous tissue simulation, as this typically occurs in a faster timescale than bone remodelling (days cf. months), and may occur before press-fit can subside. Press-fit has been seen to remain over the course of an in vivo experiment (6 weeks), when interface disruption was minimal (Ramamurti et al. 1997). Although the press-fit was seen to be reduced under the effects of excessive micromotion (Bragdon et al. 1996). The formation of fibrous tissue at the interface would, in any case, reduce the effect of press-fit, because of its greater compliance. It has been shown that bone remodelling can occur at the interface of press-fit implants in the early post-operative period (Dhert et al. 1998), but whether the press-fit persisted in this instance was unclear. Press-fit was seen to have an effect on behaviour in the fibrous tissue modelling investigation carried out previously, although this was displacement controlled. For the current investigation, it was decided to investigate the effect of press-fit by carrying out the Alloclassic simulations both with and without press-fit.

Where press-fit was required it was modelled by enforcing a uniform overclosure between the implant and bone, and then resolving this gradually over the first step of the analysis. The overclosure generated in this way negated the need for additional geometry to be created for the press-fit case, as would be required using the standard shrink-fit methods employed in the previous canine modelling investigation. The press-fit specified in this investigation was the same as that used in the canine modelling investigation (i.e. 0.1 mm radial interference). This value was used as there was no other specific information available regarding the physical level of press-fit of this particular implant, and it was thought to be a reasonable estimate.

The CAD model of the bone was the same as used in the previous design investigation. The model was truncated such that minimal analysis cost was incurred, and that the resection level did not unduly affect the results, due to proximity of imposed boundary conditions to areas of interest.

3.8.2.2 Meshing

It was noted previously that for best performance and accuracy, the finest second order tetrahedral mesh possible should be used. For the current study, the finest mesh possible was used within system limitations. This led to a typical mesh refinement of approximately 3mm. The meshed models each contained approximately 40,000 ten noded tetrahedrons. C3D10M elements were used in all cases because of the their enhanced ability in contact situations. These elements were also used in the unimplanted case to enable comparability with the other models.

Due to the inhomogeneous nature of the material properties in bone and the relatively rapid changes in density which can occur, it was desired that spatially identical meshes were available in both the unimplanted and the implanted bone models, for the bone remodelling simulation. This would prevent any initial errors caused by comparison of signals from different areas and densities of bone. This approach would require a boundary along the cut face where the implant would be seated, which would also be present in the unimplanted model. Thus for the unimplanted model, two meshed regions would be required with compatibility along this interface.

This compatibility could be achieved in either of two ways. The first and most simple method is to impose tied constraints along the boundary between the two regions. It was advised that this was not strict compatibility (ABAQUS 2002), and subsequent tests, of a similar configuration to the meshes proposed, showed that slight discontinuities could occur across such constraints, and as such this method was not used where compatible meshes were required. The second and more preferable method was to create coincident meshes at the mating surfaces, without duplicate nodes. This is typically a more difficult and restrictive meshing operation, but avoids the problems associated with tied constraints. Unfortunately, coincident meshing was not possible with Abaqus/CAE 6.3, so suitable meshes were created using the Unigraphics meshing tool, and then this mesh was converted manually to the Abaqus format. During later modelling work using Abaqus/CAE 6.4 compatible meshes were able to be created using an author developed workaround, utilising additional CUT/MERGE part tools in CAE and bespoke cutting solids created in UnigraphicsNX, to generate the necessary partition boundaries.

The non-linear geometrical solver was used in all cases because use of the linear solver was shown to lead to second order errors in required output data. Non-linear solver methods were also required to enable proper contact formulation in the implanted model. Symmetric matrix storage was used in the analysis due to system limitations, although unsymmetric matrix storage would have been preferable due to the relatively high friction analyses, which can cause unsymmetric terms.

3.8.2.3 Material properties

Similar materials properties were used as described in the previous design investigation. The bone's materials properties were assigned to the meshed model using a programme previously developed by other researchers (Taylor et al. 2002). Each element was assigned a density and appropriate materials properties based on CT scan data. The properties of the bone produced in this manner were well validated in a previous study (Taylor et al. 2002). The Wagner resurfacing implant was modelled as Cobalt Chrome alloy ($E = 220$ GPa, $\nu = 0.3$), and the cement was modelled as PMMA ($E = 2600$ MPa, $\nu = 0.3$). The Alloclassic implant was modelled as Titanium alloy ($E = 110$ GPa, $\nu = 0.33$)

3.8.2.4 Interface characteristics

Similar interface characteristics were used as described in the previous design investigation. The bone/implant interface was modelled with a friction coefficient of 0.4 and elastic slip of 20 μ m, based on information from studies by Shirazi-Adl et al. (1993) and Kuiper and Huiskes (1996). The cement/implant interface was modelled with coulomb friction of 0.3 based on information from a study by Mann et al. (1991). A small sliding contact formulation was used, as the relative motion between the interfaces was significantly less than a typical element length. The interface between cement and bone was assumed to be tied.

3.8.2.5 Loading

The muscle loading and hip joint contact loads were implemented as described in the previous design investigation. The loads were based on the data calculated by Duda et al. (1998). The muscle loads were distributed over muscle attachment areas. For the natural joint the hip joint load was distribution over a circular area of radius 17mm, derived from an in vitro study of the contact stress distributions in the human hip (Brown and Shaw 1982). The base of the bone was rigidly constrained throughout the simulation. An initial displacement was used to bring the contact surfaces together at the beginning of the simulation, to minimise chattering, although this was removed gradually before the remodelling simulation proper was started. The Wagner resurfacing prosthesis under investigation was a cemented metal component articulating on a thin polyethylene liner. A recent study by Jin et al. (1999) analysed various UHMWPE cup and rigid femoral head combinations up to 32mm diameter, under various loads up to 2500N. It was found that the half contact angle was typically between 40° and 50°. It was found in a later study by Jin (2000) that the assumption of rigidity had a negligible effect on the results. Hence, for this investigation a half contact angle of 45° was assumed for the cemented Wagner models. The load was distributed over a circular contact area with this specification.

Available clinical evidence for the Alloclassic was typically based on components which had a 28mm ball head with corresponding acetabular components, which are metal backed with a modular polyethylene articulating surface. Such a type of articulating combination has been investigated by Jin (2000). He developed a contact mechanics model based on a 28mm ball head and a 7mm UHMWPE liner, with a radial component clearance of 0.25mm, under a load of 2500N. This analysis indicated a contact radius of 9.3mm. This was implemented in the current model as a circular contact area on the ball head, such that the load vector was coincident with the surface normal at the joint contact area centroid. The load was applied to nodes within the contact area of the meshed model. Loads were applied in the direction of the required vector, rather than as a pressure normal to the surface.

3.8.2.6 *Abaqus user-subroutines*

The predictive simulations relied upon the additional functionality available from user subroutines which interacted with the FE analyses. The prime function of these subroutines was alteration of the material properties over multiple steps during the analyses, using the remodelling rules. The subroutines which were written for bone remodelling and fibrous tissue simulations are detailed in Appendix A. The main user-subroutine functions were:

- UVARM – output the required information from the unimplanted model which was used as the reference for the changes which occur due to the implantation.
- ORIENT/SDVINI – assigned initial bone materials properties and reference signal levels for the implanted model based on information from the unimplanted bone.
- USDFLD – updated the values of density/tissue type (and hence material properties) at each material point, at the beginning of each remodelling load step, based on the values of signal determined at the end of the previous step using the remodelling rules.

The remodelling rules were different for the bone remodelling and the fibrous tissue simulations, but were both based on theories which relate the biological changes in the material properties over time to mechanical signals sensed locally. Both methods could be applied on the same model, and the combined effects could also be investigated as required.

3.8.2.7 *Bone remodelling algorithm*

The bone remodelling algorithm was based on a method used in previous investigations to simulate the changes in bone density which occur in the femur after implantation of a prosthesis (Schmitz et al. 2004). This method was chosen as there was significant previous experience of using this method, and it had produced some clinically relevant results in previous investigations (Taylor et al. 2004, Schmitz et al. 2004).

The algorithm compares the levels of Strain Energy Density (SED) before and after implantation, and changes the bone material properties based on the difference in local values. If the SED increases then an increase in density is prescribed, and conversely if the implantation causes a decrease in SED then a local decrease in bone density is prescribed. A schematic representation of the algorithm is shown in Figure 35. The actual amounts of bone density change are controlled by a number of factors which are detailed here.

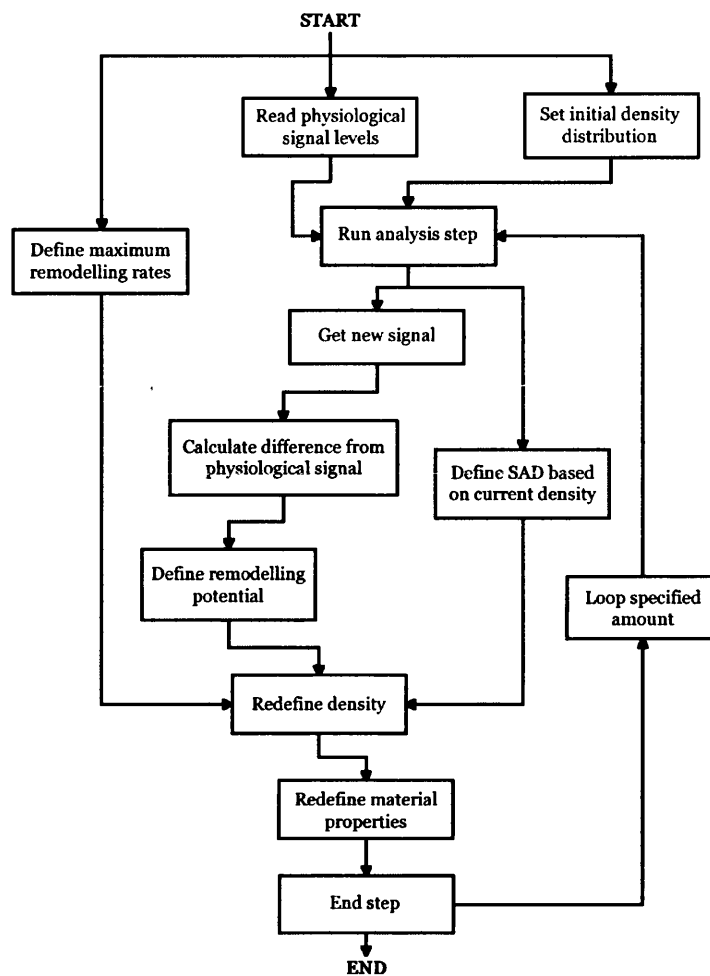


Figure 35: Process flow diagram for bone remodelling algorithm

For the analysis, physiological loading was applied over multiple steps to simulate 10 years bone remodelling. The parameters were chosen to represent 6 month periods for each step, in order to be able to complete the simulation in reasonable length of time (approximately 2-3 days). Other parameters were used to control the local rates of remodelling. A parameter was required in the definition of the empirical remodelling curve (Figure 36), such that a significant change in SED was required before any bone density change can occur. This parameter is sometimes referred to as the lazy zone. Values for this parameter that have shown reasonable bone remodelling behaviour have been between 35% and 75% of the physiological level of SED (Huiskes and van Rietbergen 1995). Practically this parameter manifests itself as the level of reactivity of the bone, where a level of 35% is more reactive than a level of 75%. It is thought that this level of reactivity can vary throughout the population and between species. However, it was shown by Huiskes that changing the value of the lazy zone width had an effect on the amount of bone remodelling, but did not have a significant effect on the pattern of remodelling (Huiskes et al. 1992). The lazy zone was implemented in this study such that a change in the SED of 35% or more from the physiological normal value was required before any

remodelling could occur. Another boundary value was imposed such that a change of 100% or more from the physiological value led to the maximum possible remodelling rate.

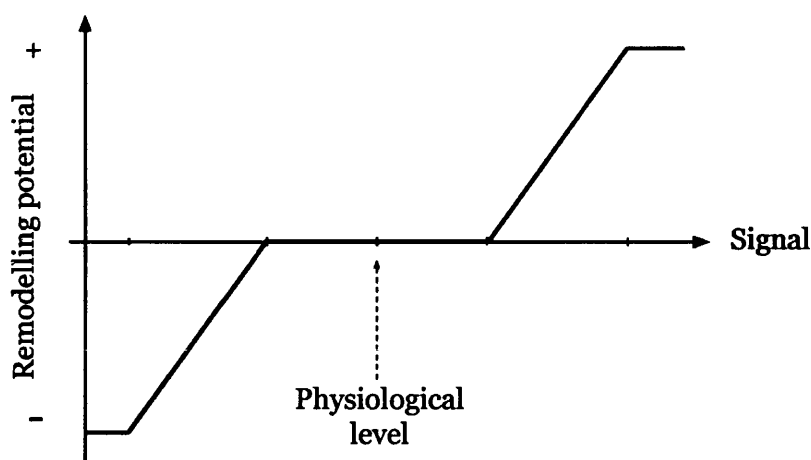


Figure 36: Remodelling curve

Actual remodelling rates were required to be able to link the changes in density to the timescales over which they would occur. Data was used from bed rest studies (LeBlanc et al. 1990). The region of maximum femoral bone loss during the study was used, as worst case loss was the required parameter. The trochanter showed a 4.6% loss over the 17 week bed rest period. The remodelling theory used here required an actual mass change rather than a percentage change in density. To relate this percentage change to meaningful values a sample of the trochanter of the cadaveric model through the antero-posterior viewing direction was taken. The percentage change was thus assumed to apply to the average density of this sample. This led to a maximum loss rate of 29.6 mg/cm³ per 6 month step.

The effect of surface area density (SAD) was included in the model as in previous remodelling simulations. SAD was introduced as it was recognised that different densities of bone have different rates of remodelling, dependant on the bone surface area. The function defining the relationship between SAD and density of bone was based on a function used by (Beaupré et al. 1990). The direct relation between the effect of SAD and the bone specific deposition/resorption rates was based on the previous sample of the trochanter of the bone model, and an average value of SAD was used, weighted by volume. The remodelling rates based on the trochanter region, were thus applied for this average value of SAD. The remodelling rates were made proportional to the SAD which varied with the density of the bone. During the analysis, the bone density was prevented from exceeding physiologically meaningful levels (0-2 g/cm³).

3.8.2.8 Fibrous tissue algorithm

The fibrous tissue algorithm was essentially the same as the bone remodelling algorithm, in that the levels of signal generated in the tissue caused changes in material properties to occur. The algorithm was based on the work of Büchler et al. (2003), whereby the level of cyclic octahedral shear strain governed the differentiation of bone into fibrous tissue. The method was adapted however, to take account of the significant shear strains which occur during physiological loading of the intact bone. If left unmodified, the method would indicate fibrous tissue formation in the natural unimplanted bone, which is not realistic. A schematic diagram of the remodelling algorithm used in this investigation is shown in Figure 37.

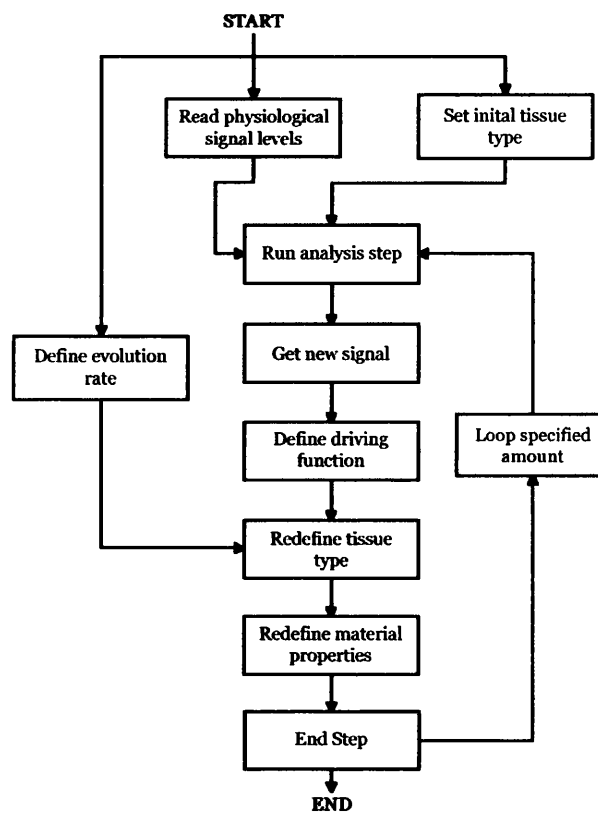


Figure 37: Process flow diagram for fibrous tissue simulation

The tissue differentiation was controlled by the level of cyclic octahedral shear strain, relative to that seen in the physiological case which would govern whether bone or fibrous tissue was formed. Figure 38 shows the tissue differentiation driving function, which determined the tissue type based on the mechanical stimulus, E . The stimulus used for the driving function was the cyclic octahedral shear strain. The driving function, $M(E)$, was such that $M(E)=0$ led to bone formation, and $M(E)=1$ led to fibrous tissue formation. The parameters E_{min} and E_{max} corresponded to the thresholds of micromotion $20\mu\text{m}$ and $150\mu\text{m}$ respectively. These parameters were obtained by Büchler et al. (2003) based on the in vivo canine experimental studies of Jasty et al. (1997) and the parameters were varied

until similar behaviour to the in vivo experimental data was obtained. This led to values of $E_{min} = 2.3 \times 10^{-3}$ and $E_{max} = 8.2 \times 10^{-3}$.

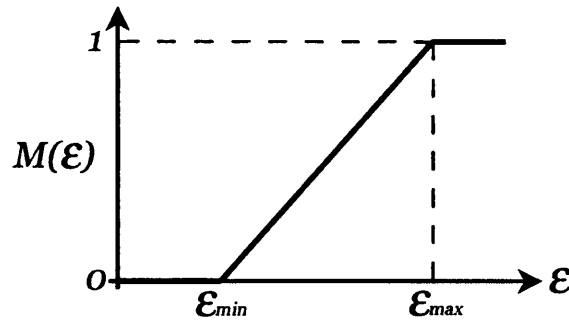


Figure 38: Fibrous tissue driving function

Due to the significant levels of cyclic shear strain seen in the unimplanted bone, the cyclic shear strain thresholds were assumed to be absolute - over and above the physiological levels. This provided a local effect, similar in nature to the bone remodelling algorithm. As was stated previously, the signal was investigated in two forms: The cyclic octahedral shear strain, which did not include the component produced by press-fit; and the total octahedral shear strain, which was inclusive of the press-fit component. These two methods were both tested on the clinical cases investigated, and the most clinically relevant method was pursued.

The fibrous tissue was allowed to follow a global rule, such that it could form anywhere in the bone if the threshold signals were exceeded. Fibrous tissue simulation was carried out over a series of 21 steps, which was ostensibly equivalent to 21 days for fibrous tissue formation, using rate parameters derived by Büchler et al. (2003). Loading which included press-fit required an extra relaxation step for the cyclic method, to take account of the strains when the implant was not being directly loaded. This practically doubled the total analysis time, and so only a maximum of 10 fibrous tissue remodelling steps could be simulated within a reasonable timescale, although it was thought that this could still provide enough information on the initial fibrous tissue behaviour.

3.8.3 Results

To assess the effects of the different predictive simulations, the results of the bone remodelling and fibrous tissue simulations were examined individually for the implants investigated. Then the combined effect of the simulations on each other was investigated, along with other findings of interest. The simulations were examined for their clinical relevance.

3.8.3.1 Bone remodelling simulation results

To visualise the main effects of bone remodelling, cross-sections were typically taken through the bone perpendicular to the antero-posterior viewing plane for each implant, as this is typically the view available in clinical radiographic follow-up. This is also the plane in which most changes can typically be seen as it is perpendicular to the main loadpath.

Wagner (Symmetric)

The final pattern of bone density after 10 years simulated remodelling is shown in Figure 39a. This indicates a denser central bony column, with areas of lower density bone either side. The absolute change in density (Figure 39b) indicates that a large area under the superior aspect of the cup has typically lost around 200 mg/cm³ over the course of the simulation. The density change as a percentage of the original bone density is shown in Figure 39c. This indicates around 50% bone density losses, either side of the central region. The progression of the bone remodelling is shown over the course of the simulation in Figure 39d. This indicates that the density decreases first on the medial side and then on the lateral side.

Wagner (Valgus)

The final pattern of bone density after the remodelling is shown in Figure 40a for the Wagner (Valgus). This also indicates a central column of bone, but this is not as distinct as with the Wagner (Symmetric), with more bone loss on the lateral side. The absolute change in bone density (Figure 40b) shows a significantly larger area of bone loss under the cup, of which a large proportion is over 200 mg/cm³ of density loss. The percentage change of bone density (Figure 40c) indicates losses typically over 60%, which were located mainly on the lateral side of the implant.

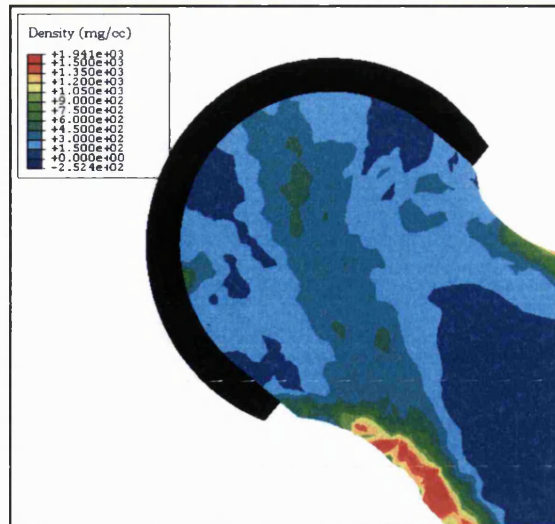


Figure 39a: Final bone density for Wagner (Symmetric)

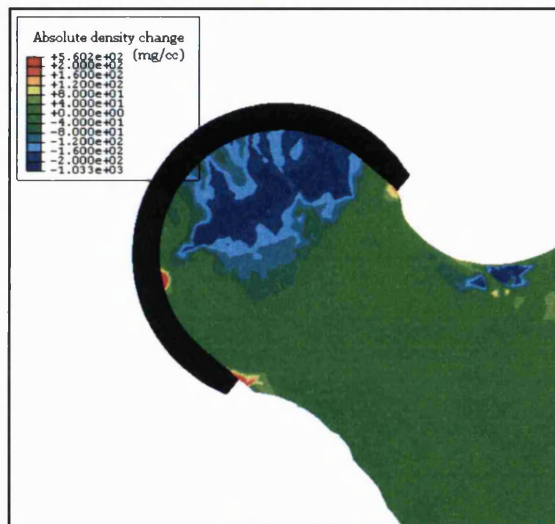


Figure 39b: Absolute change in bone density for Wagner (Symmetric)

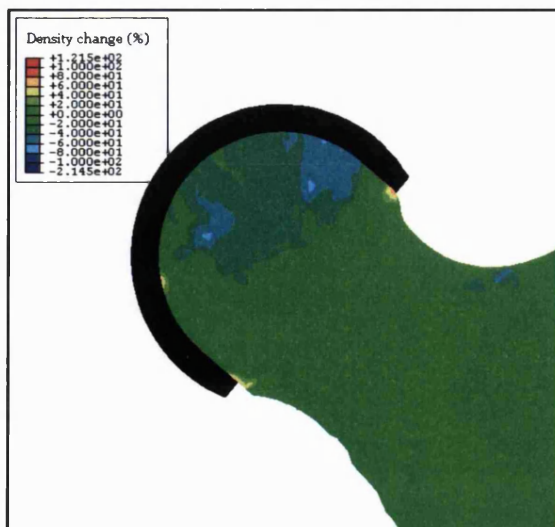


Figure 39c: Percentage change in bone density for Wagner (Symmetric)

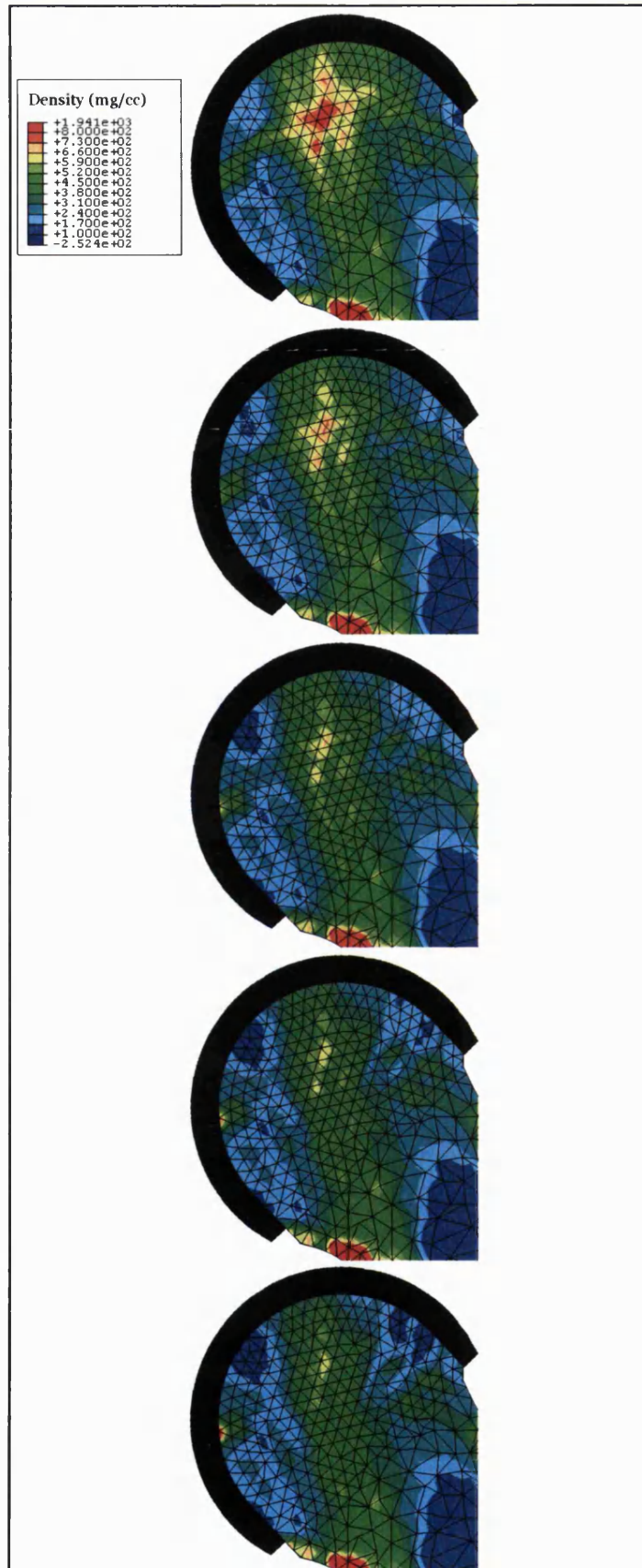


Figure 39d: Progression of bone density changes for Wagner (Symmetric)

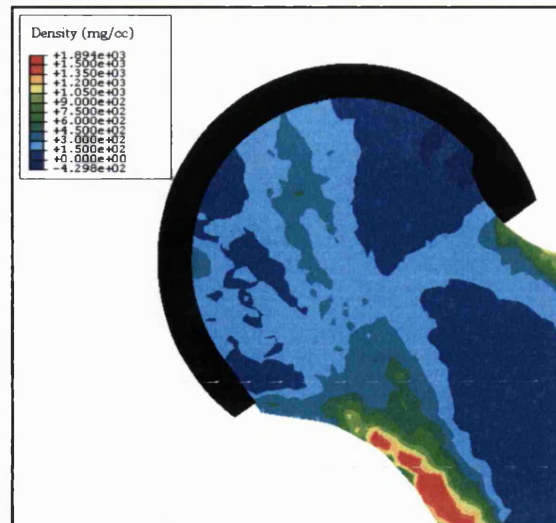


Figure 40a: Final bone density for Wagner (Valgus)

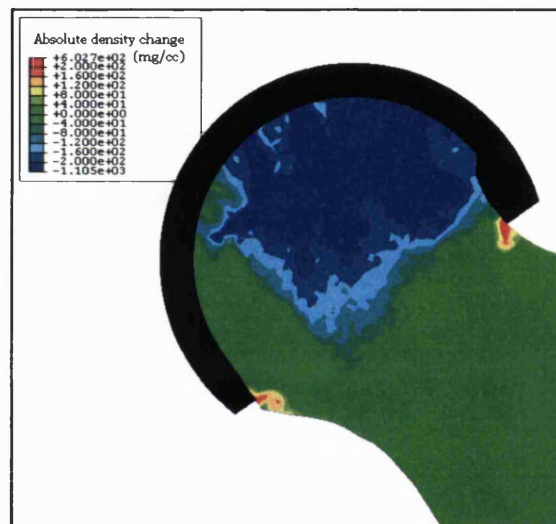


Figure 40b: Absolute change in bone density for Wagner (Valgus)

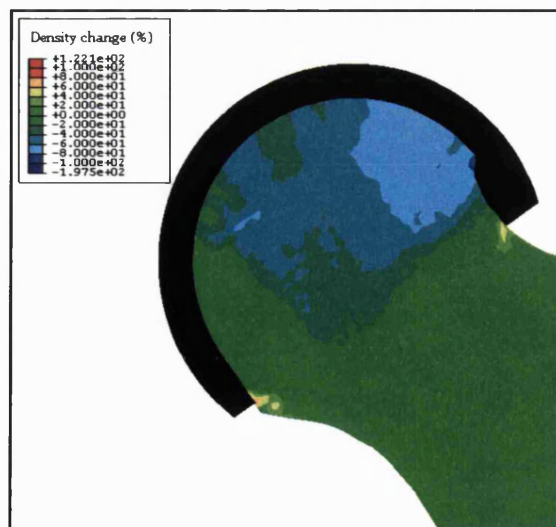


Figure 40c: Percentage change in bone density for Wagner (Valgus)

Alloclassic

The absolute change in density over the course of the 10 year bone remodelling simulation is shown in Figure 41a, for both the press-fit and the no press-fit models. In the case without press-fit it can be seen that there is significant bone loss over almost the entirety of the length of the stem, especially in the cortical bone regions. These losses are typically well over 200 mg/cm³. For the case with press-fit there are also some losses in the cortical bone, but the losses are confined to the more proximal regions and are less extensive. There is also evidence of increases in bone density around the distal stem, in the press-fit case.

Looking at the percentage changes in bone density over the course of the experiment (Figure 41b) it can be seen that the majority of the bone density changes in the non-press-fit case are between 40% and 80% loss of the original density over the length of the stem. In the case with press-fit a far smaller area of bone loss is apparent. This is restricted mainly to the proximo-medial cortical bone where there are also typically losses of 40% to 80% density. This area is immediately adjacent to an area of density increases of 60% to 80% bone density are seen in the bone close to the implant. A similar pattern is seen on the lateral side to the medial side, but typically with less losses in evidence. In both cases (with and without press-fit) there is some evidence of density increases in the cancellous bone at the very tip of the implant. This is more apparent with the press-fit case.

In the 3D views (Figure 41c and 41d) it can be seen that the losses predicted for the non-press-fit case are far more extensive than those predicted for the case with press-fit. In the non-press-fit case the losses are seen to reach over 80% and are spread over the majority of the upper femur. In the press-fit case the losses are less severe and restricted to the proximo-medial region with some in the proximo-anterior region. There are also some density gains evident in the remainder of the femur.

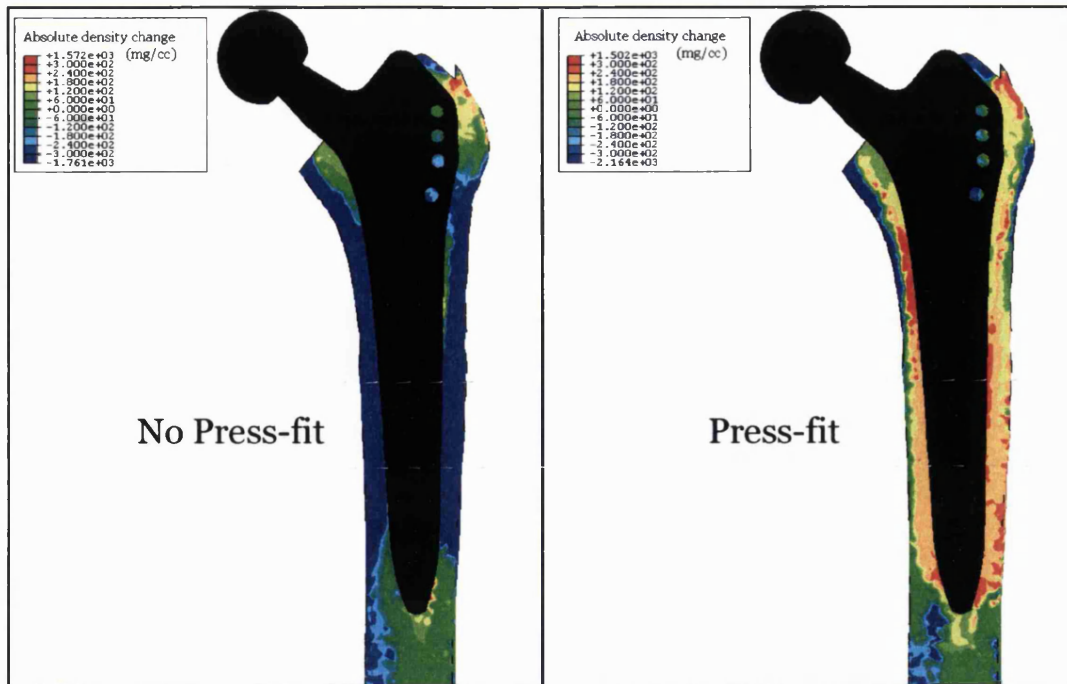


Figure 41a: Absolute change in bone density for Alloclassic

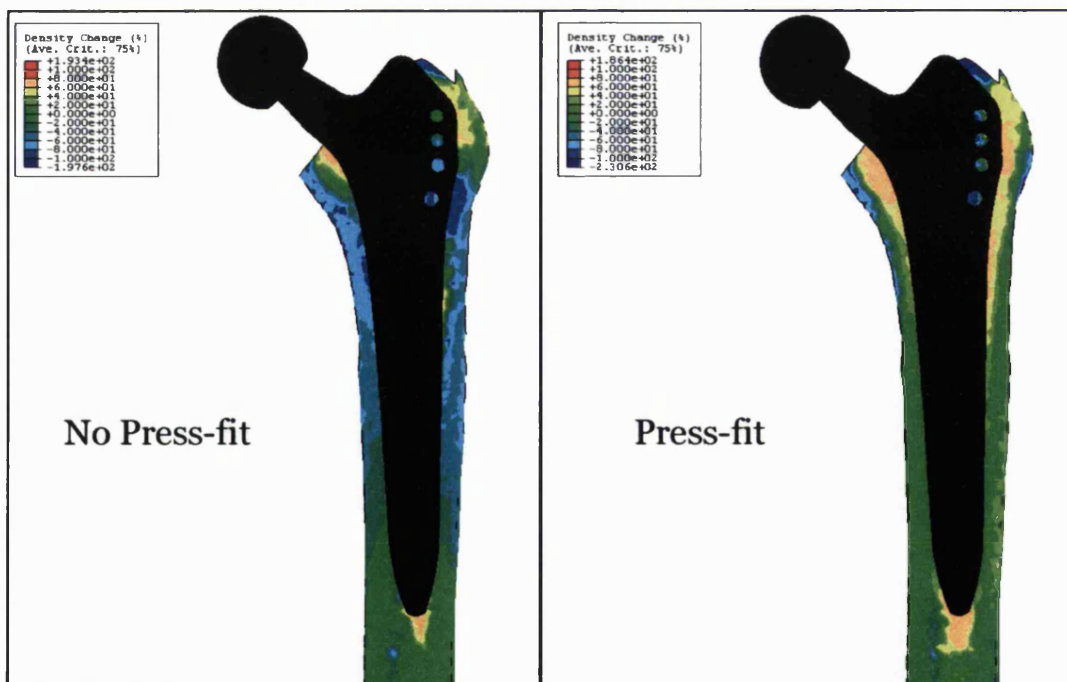


Figure 41b: Percentage change in bone density for Alloclassic

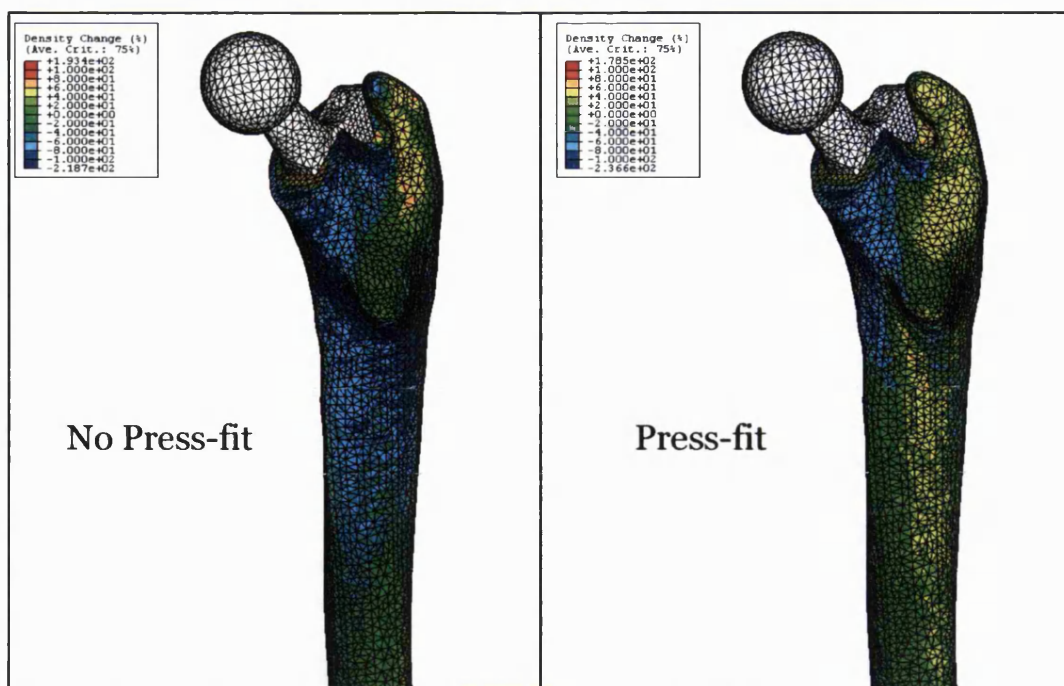


Figure 41c: Percentage change in bone density for Alloclassic (posterior 3D view)

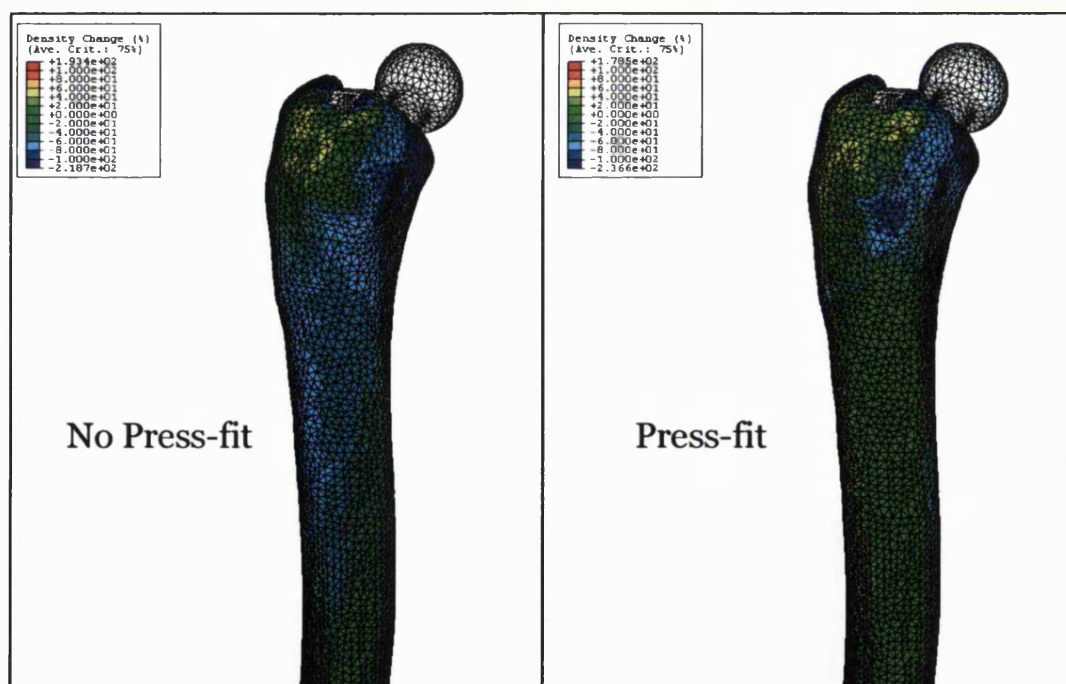


Figure 41d: Percentage change in bone density for Alloclassic (anterior 3D view)

3.8.3.2 Clinical comparison with bone remodelling results

The results of bone remodelling on the Wagner indicate that the simulation can produce clinically relevant results. Earlier investigations found that a typical clinical pattern of bone remodelling involved diffuse bone loss in the head, combined with a central bony column typically with increased density under the apex of the cup. This is similar in nature to results obtained from the bone remodelling simulation here. The simulation results did indicate a general decrease in bone density under the cup, leading to a central bony column, although no sclerosis was noted. The results of the Wagner simulations also indicate that the orientation/size of the component can affect the results.

The results of the Alloclassic with press-fit are very similar to clinical behaviour. Earlier investigations found that a typical clinical pattern of bone remodelling involved bone atrophy proximally in Gruen zones 1 and 7, and evidence of cortical hypertrophy and thickening distally around the stem, with some cases which exhibited pedestal formation at the tip of the stem. This is very similar behaviour to the bone remodelling simulations of the Alloclassic with press-fit.

The Alloclassic without press-fit produced results inconsistent with the clinically observed patterns of bone remodelling. The simulation without press-fit predicted massive amounts of bone loss along the majority of the stem length, which would be ultimately devastating for the bone in question, leading to failure which would require extensive revision surgery. This is contrary to the significant amount of long-term clinical success seen with this implant.

3.8.3.3 Fibrous tissue controlling signal investigation

At the end of the previous interface modelling investigation, there were two proposed methods of fibrous tissue formation (using Cyclic and Total octahedral shear strain as a signal). In order to discern the better method it was required that both methods should be implemented on a clinically relevant case. Both methods were initially tested on the Wagner and the Alloclassic.

Of the two Alloclassic models, the Alloclassic with press-fit was thought to be the most appropriate, because the bone remodelling simulations for this case were seen to produce more clinically relevant results. Notwithstanding this, the press-fit is likely to remain over the short term, and is likely to be within the timescales of fibrous tissue formation. This is endorsed by the fact that the press-fit was seen to remain throughout the course of a 6 week fibrous tissue experiment, when the interface was not disrupted by large micromotions (Ramamurti et al. 1997).

Total octahedral shear strain method tests

Evidence was found that the total octahedral shear strain method did not produce clinically relevant results. When tested on the Wagner resurfacing prosthesis, the threshold was never exceeded, and fibrous tissue was not produced. This was found for both the symmetric and the valgus versions, both before and after 10 years bone remodelling, and with or without cement/bone interface failure modelling. This was contrary to clinical evidence of fibrous tissue formation seen in the Wagner.

Further evidence on this matter was discernible from the behaviour of the Alloclassic simulations. Using the total octahedral shear strain method with the Alloclassic (with press-fit) indicated fibrous tissue production at the midstem and tip regions (Figure 42), and not proximally. This behaviour was contrary to the available clinical evidence on the behaviour of the Alloclassic. A similar pattern of fibrous tissue formation was also apparent after 10 years bone remodelling.

Cyclic octahedral shear strain method tests

Contrary to the alternative method, the cyclic octahedral shear strain method did show behaviour more relevant to the clinical experience. When the cyclic octahedral shear strain method was used on the Wagner, fibrous tissue was produced, and this is discussed in more detail later.

Further relevant clinical behaviour was found with the Alloclassic model. The results of the cyclic octahedral shear strain method are shown for the Alloclassic (with press-fit) in Figure 43. This method indicates fibrous tissue formation in the proximal region of the implant only. This agrees with the behaviour seen clinically for the Alloclassic, where

proximal radiolucencies are typically seen in radiographs, which can be evidence of fibrous tissue at the interface.

Conclusions

The total octahedral shear strain method of fibrous tissue formation was shown not to be a good representation of clinical behaviour. Conversely, the cyclic octahedral shear strain method was shown to be a reasonable representation of clinical behaviour. Therefore, for the remainder of the predictive investigation, the total method was discarded in favour of the cyclic method.

The cyclic method is similar to that originally proposed by Büchler et al. (2003). This method now has more justification in terms of the nature of the signal used. The results of this comparison also indicate from previous discussions, that the signal to which the bone responds is likely to be related to the cyclic component of the signal only, such as fluid flow. This is in contrast to the total signal, which would have indicated that damage based signals would be more likely.

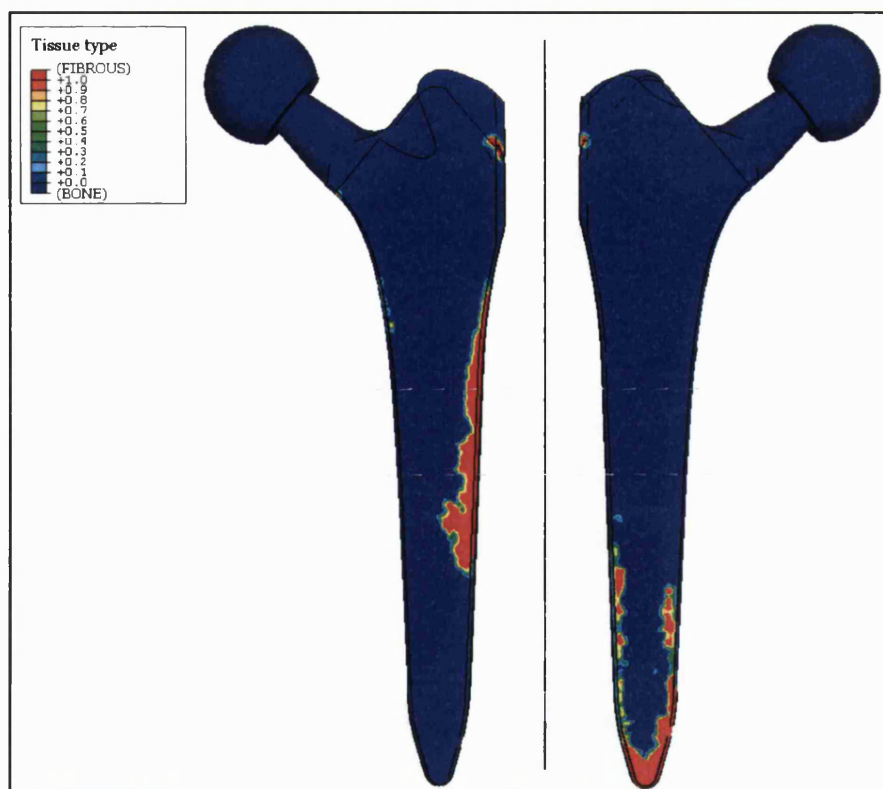


Figure 42: Fibrous tissue formed by Total octahedral shear strain method for Alloclassic (with press-fit)

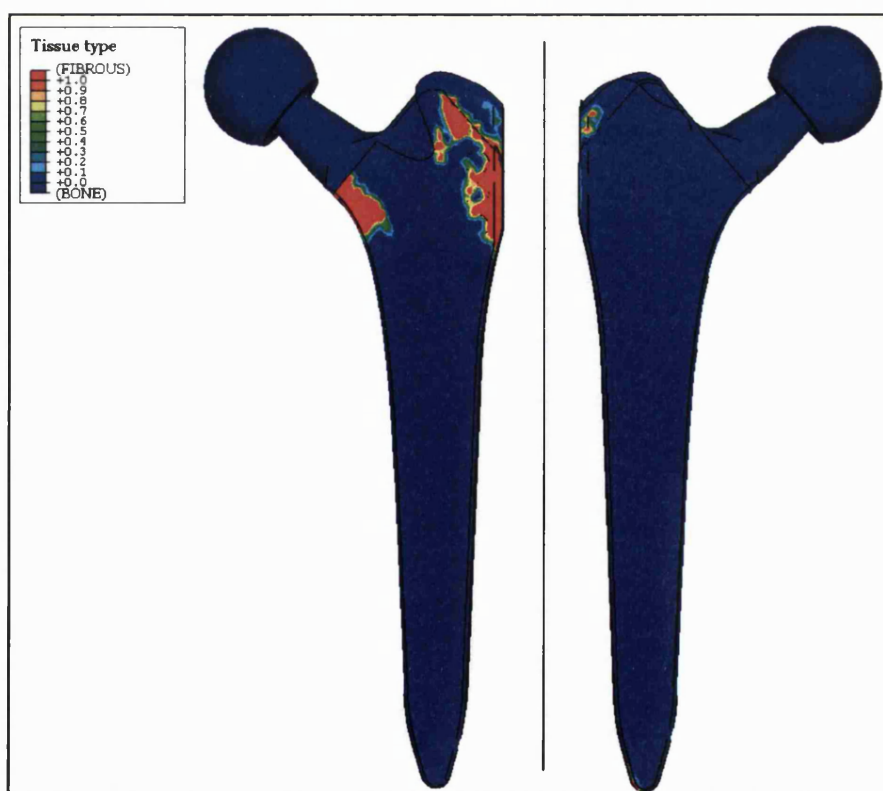


Figure 43: Fibrous tissue formed by Cyclic octahedral shear strain method for Alloclassic (with press-fit)

3.8.3.4 Fibrous tissue simulation results

The patterns of fibrous tissue over the course of the simulation are detailed here. Patterns are typically shown at the end of the simulation. To visualise the fibrous tissue formed at the interface, implants are typically removed from the bone for clarity of viewing. In the following diagrams tissue type is indicated by colour. Fibrous tissue is indicated by the colour red and bone is indicated by blue, with a mixture of the two tissues indicated by the intermediate colours.

Wagner (Symmetric)

In this simulation (Figure 44a) fibrous tissue was seen to occur initially around the edge of the implant. Then an increase in fibrous tissue was seen to occur on the medial side. This bloom of fibrous tissue on the medial side then increased in size (Figure 44b) and traversed the neck in the plane of the prosthesis. The analysis aborted after 17 remodelling steps (a simulated 17 days) due to excessive deformations in the elements. A large proportion of the neck was occupied by fibrous tissue at this point, and this significantly weakened the structure, causing collapse of the head under load - which lead to excessive element distortion, and failure of the analysis.



Figure 44a: Fibrous tissue formed with Wagner (Symmetric)

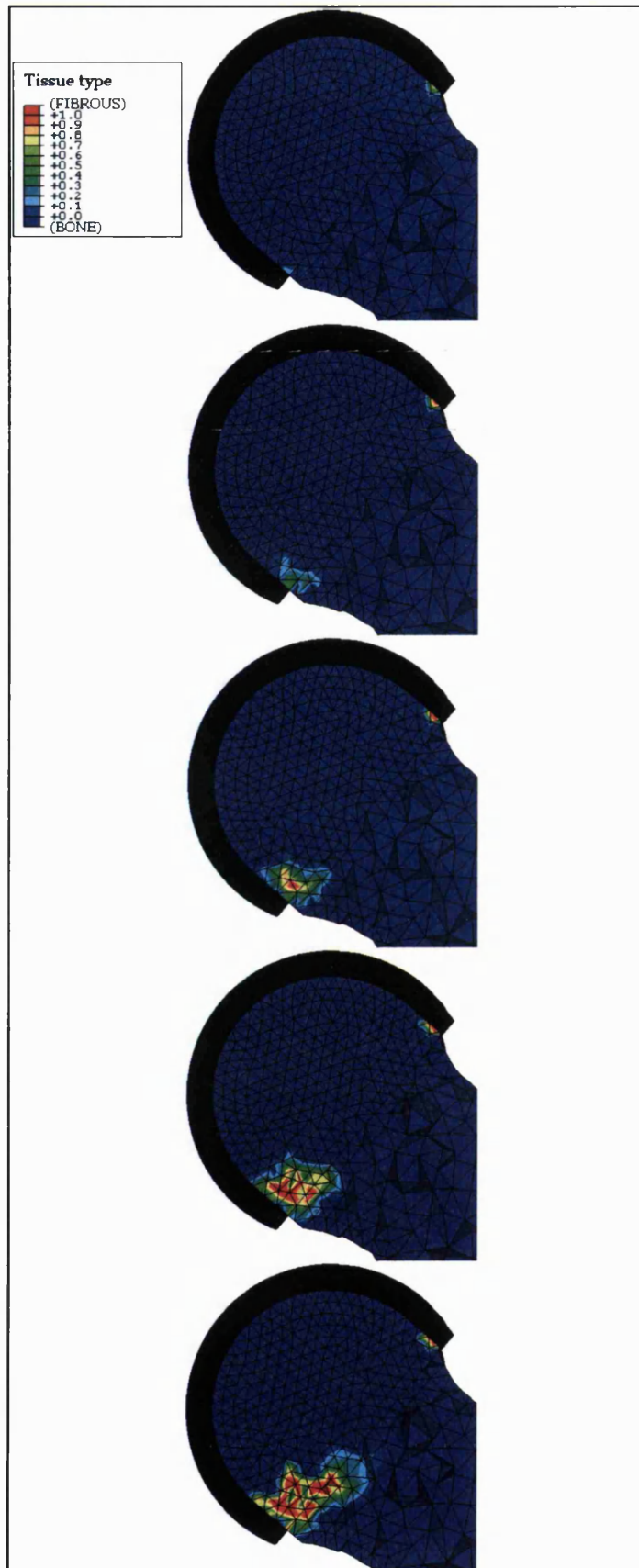


Figure 44b: Progression of fibrous tissue formed with Wagner (Symmetric)

Wagner (Valgus)

The Wagner (Valgus) behaved in a slightly different manner to the symmetric version. The fibrous tissue initially formed around the edge of the implant as before, and then the tissue concentrically increased in depth across the plane of the prosthesis, but the majority of the tissue was on the anterior side instead of the medial (Figure 45). The progression of the analysis was slower than the symmetric version, and completed all 21 steps (3 simulated weeks) that were prescribed.

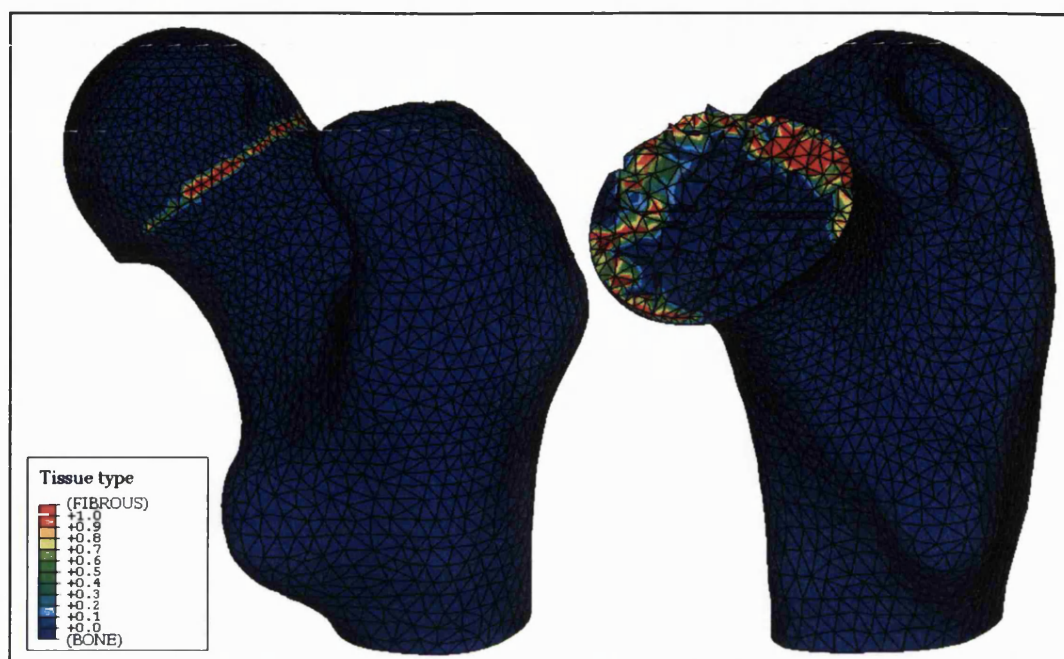


Figure 45: Fibrous tissue formed with Wagner (Valgus)

Alloclassic (without press-fit)

The fibrous tissue simulation for the Alloclassic without press-fit is shown in Figure 46. This indicated a significant amount of fibrous tissue extending from the proximal region to the midstem, on the medial and lateral sides, and extending onto the posterior side of the implant. A small area of fibrous tissue was also formed at the tip of the implant. The areas of fibrous tissue were large and progressed rapidly from proximal to distal. The analysis aborted after 6 fibrous tissue remodelling steps (simulated 6 days) due to excessive deformations in the elements.

Alloclassic (with press-fit)

The results for the Alloclassic with press-fit are shown in Figure 47. This indicated less extensive fibrous tissue than the case without press-fit. The fibrous tissue for this case was also indicated in the proximal areas on the medial and lateral sides. The progression of the fibrous tissue was much slower and appeared to stabilise in some areas. The analysis completed all of the 10 fibrous tissue remodelling steps which were prescribed. A comparison of the rate of progression of the two analyses (without and with press-fit), at

similar time-frames can be seen in Figure 48. It can be seen that the fibrous tissue produced by the case without press-fit was far more extensive and progressed significantly quicker, than the case which included press-fit.

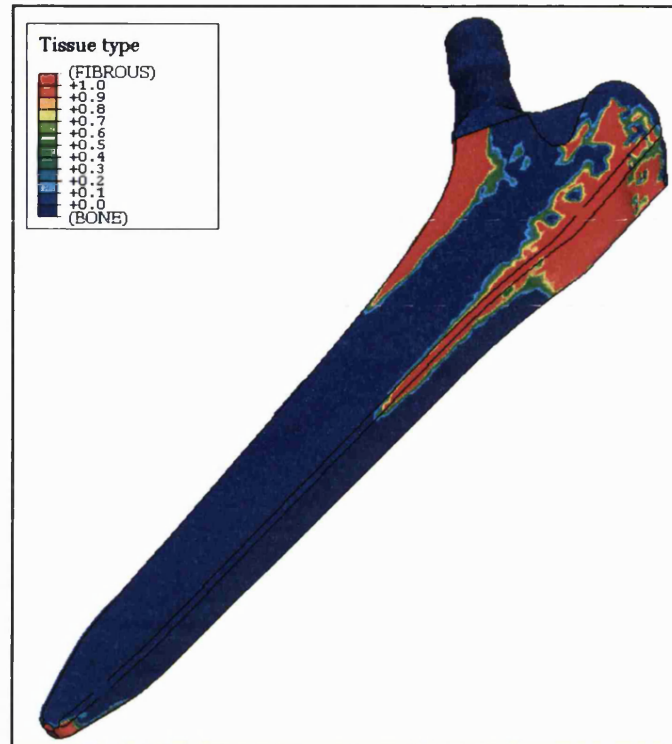


Figure 46: Fibrous tissue formed after 6 steps with Alloclassic (no press-fit)

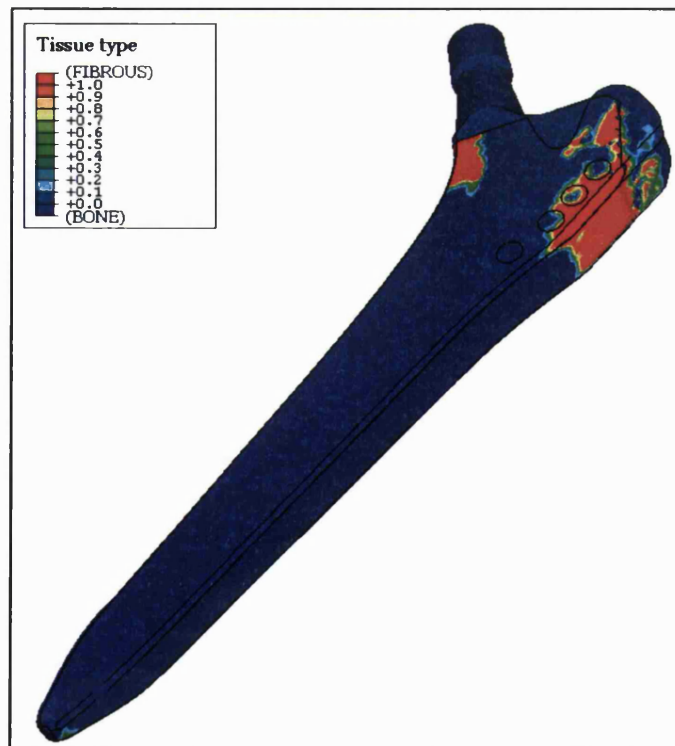


Figure 47: Fibrous tissue formed after 10 steps with Alloclassic (with press-fit)

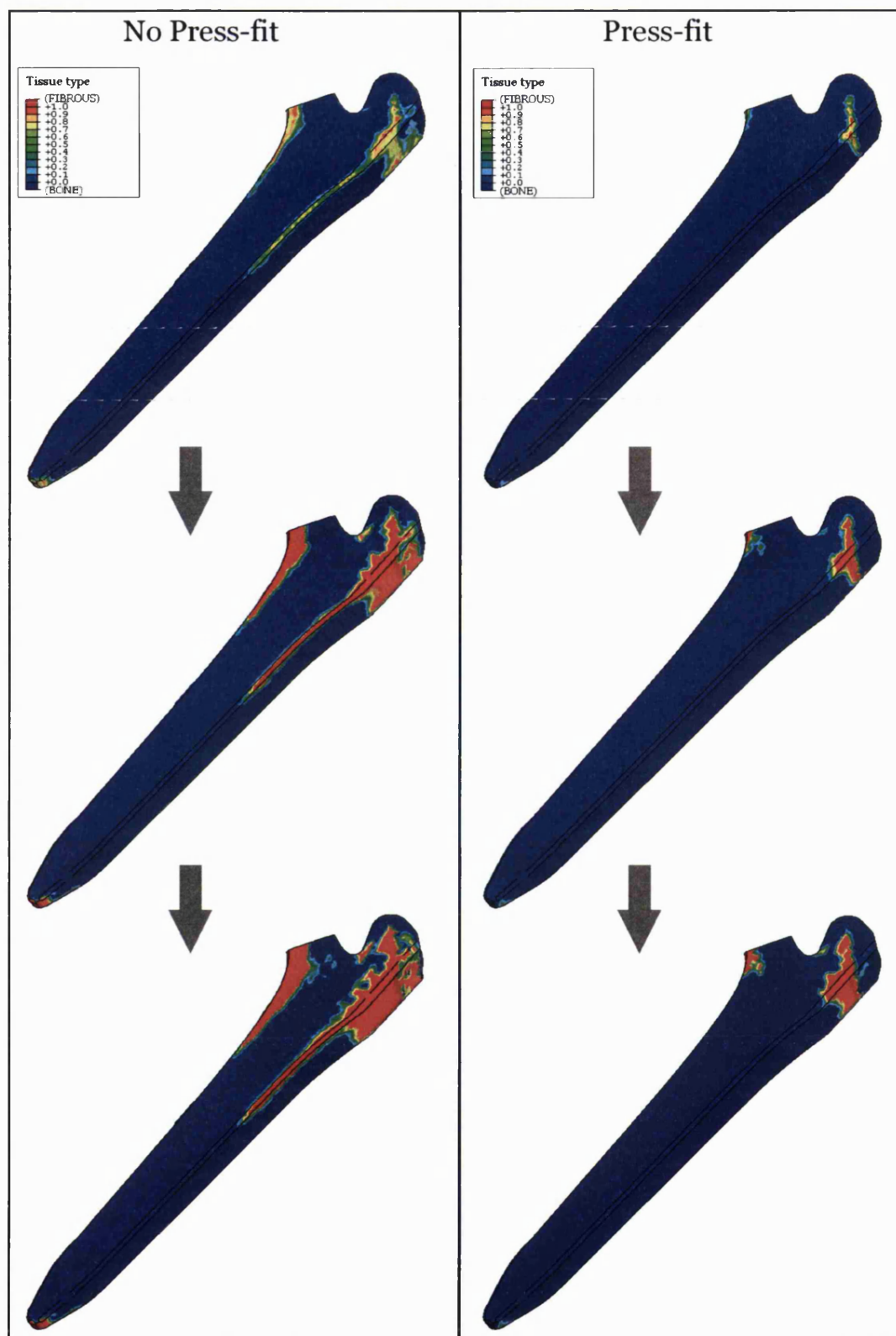


Figure 48: Effect of press-fit on fibrous tissue progression with Alloclassic

3.8.3.5 Clinical comparison with fibrous tissue results

Clinical evidence for failed Wagner prostheses has suggested that typically, fibrous tissue initiated at the periphery of the cup and advanced towards the apex. This appeared to start at the medial edge and progress to the lateral side, eventually leading to gross loosening and clinical failure.

The Wagner fibrous tissue simulations showed similar behaviour to this, where fibrous tissue was initially formed at the periphery of the cup. The progression of the fibrous tissue is then slightly different. In the simulation, the tissue front does not advance significantly towards the apex, but tends to traverse the neck of the femur. Some proximal movement can be discerned in the bloom which occurs on the medial side. And the fibrous tissue is more dominant on the medial side, as seen in clinical behaviour.

Clinical evidence for the Alloclassic has typically shown radiolucent lines proximally in Gruen zones 1 and 7. This can be evidence of fibrous tissue formation at the interface (although periosteal bone remodelling has also been seen to cause this). These radiological symptoms exhibited do not appear to have a significant effect on the longevity of the implant, which has very good long-term results.

The Alloclassic with press-fit produced a similar pattern to those seen radiologically, as fibrous tissue was formed in the more proximal regions on the medial and lateral side of the implant, with no significant fibrous tissue forming at the midstem or distal regions.

The simulation of Alloclassic without press-fit exhibited extensive fibrous tissue formation in the proximal regions which extended to the midstem region, combined with some tissue at the tip of the stem. This simulation exhibited more fibrous tissue, extending further distally than is typically seen in the clinical situation, and with additional tissue at the tip.

3.8.3.6 Combined effects of simulations

The combined effects of the bone remodelling simulation and the fibrous tissue simulation are obviously of interest, with respect to the effect that they have on each other. It should be borne in mind that the relative timescales of tissue transition are significantly different between the two biological effects. Fibrous tissue is typically formed over a matter of days and weeks, whereas significant bone remodelling effects occur over months and years.

The effect of the presence of fibrous tissue on the bone remodelling simulation is illustrated in Figures 49a and 49b. Figure 49a shows the absolute bone changes which occurred, with and without fibrous tissue present initially. It can be seen that there are no significant changes in the overall pattern, except in the proximo-lateral cortex, where bone density losses are reduced by the presence of fibrous tissue. Figure 49b shows the percentage changes in bone density for the same cases. It can be seen from this that the areas of cancellous bone in the most proximal and distal regions which increased in

density in the absence of fibrous tissue, experience less bone density increase if fibrous tissue is present. However, it is noted that bone density increases persist immediately adjacent to the areas of fibrous tissue.

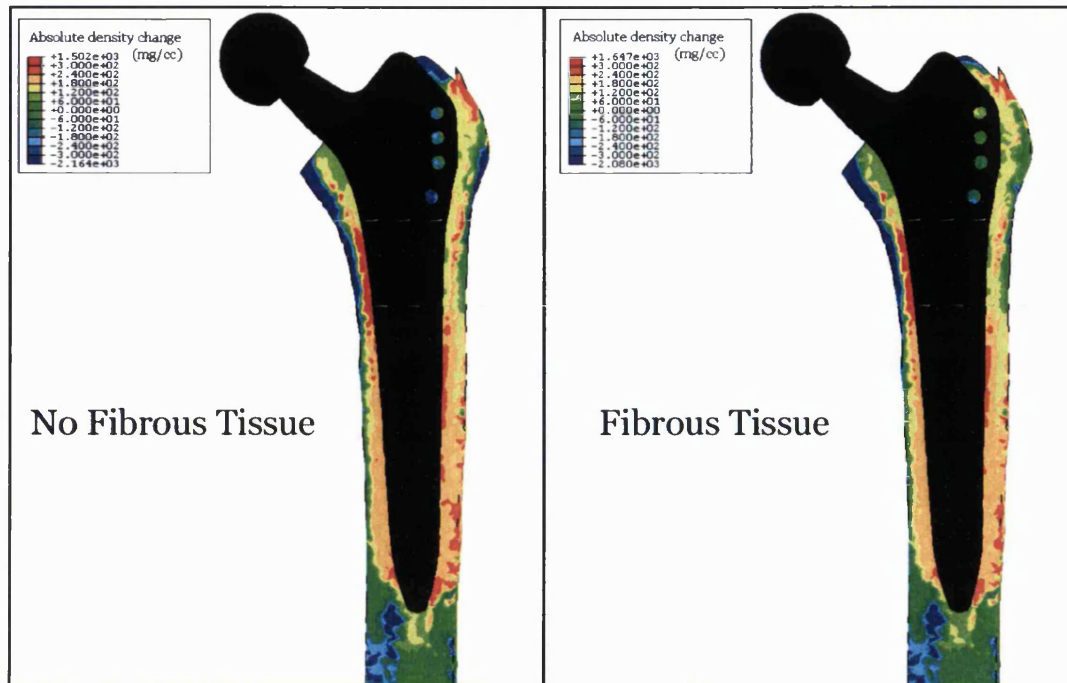


Figure 49a: Effect of fibrous tissue on absolute levels of bone remodelling for the Alloclassic (with press-fit)

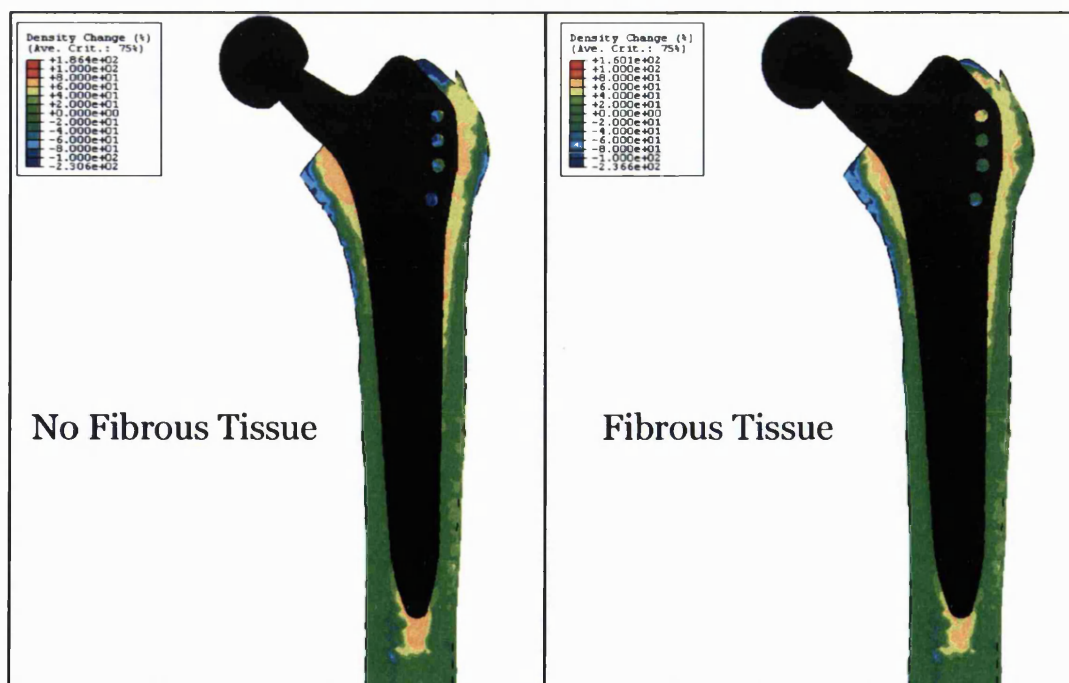


Figure 49b: Effect of fibrous tissue on percentage levels of bone remodelling for the Alloclassic (with press-fit)

The change in indications for fibrous tissue after bone remodelling has occurred are shown in Figure 50. This indicates that the effect of 10 years bone remodelling was not significant on the fibrous tissue. However, some areas of change can be seen such that small decreases in fibrous tissue area would be experienced in the proximal region after bone remodelling.

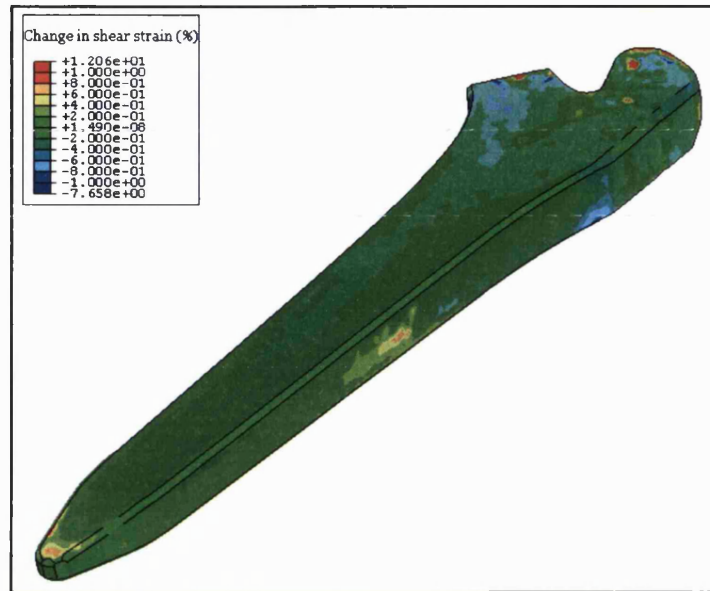


Figure 50: Percentage changes in cyclic octahedral shear strain due to 10 years remodelling (after fibrous tissue formation)

The effect these changes would have is illustrated in Figure 51. This shows that the proximo-medial area of fibrous tissue partly reverts back to bone, and fibrous tissue growth is arrested in other areas, due to the changes occurring as a result of bone remodelling.

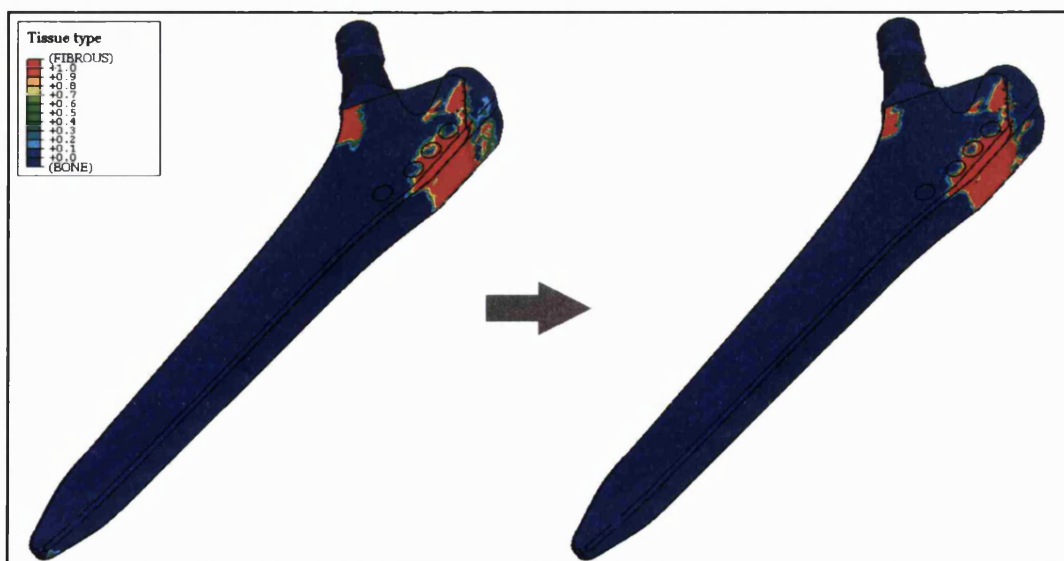


Figure 51: Change in fibrous tissue formation after 10 years bone remodelling

These combined simulations have shown that the fibrous tissue and remodelling simulations can affect each others outcomes. It was noted that bone density increases were present in the proximal regions immediately adjacent to the regions of fibrous tissue formation. This type of behaviour has also been noted during in vivo canine studies by Jasty et al. (1997). This increased bone remodelling adjacent to fibrous tissue may also be symptomatic of increased radiographic contrast of radiolucent lines seen in the longer-term during in vitro experiments (Sew Hoy et al. 1981). This behaviour in the Alloclassic combined model may be indicative of the detection of radiolucent lines proximally in the Alloclassic clinical experience.

3.8.3.7 Additional investigations/results

Detailed here are additional investigations carried out alongside the main investigations, to gain extra insight into the behaviour of the simulations.

Restricted fibrous tissue model

In the main analyses there was no restriction on which areas of bone could change into fibrous tissue. This was felt appropriate because imposing other constraints was thought to be overly prescriptive, and may interfere with the fundamental nature of the simulations. In light of the results for the fibrous tissue simulations on Wagner, the possibility of restricting the area over which fibrous tissue could form was investigated. This was because the apparent traversing of the neck with fibrous tissue, was not completely in line with clinical observations. Clinical observations commonly identified an advancing front of fibrous tissue along the bone interface, which travelled towards the apex of the cup.

For this sub-investigation it was hypothesised that fibrous tissue could only form local to the bone interface. The idea of restricting the active region of the fibrous tissue simulation to the first few millimetres of bone depth, may have some grounding in that this area may have a locally altered biomechanical response. This could be due many factors such as altered blood flow, response due to surgical trauma, or a local immune response.

To implement this hypothesis the fibrous tissue remodelling simulation was restricted to the first few millimetres of bone elements closest to the cement layer on the Wagner (Symmetric) model. The fibrous tissue simulation was then run as normal.

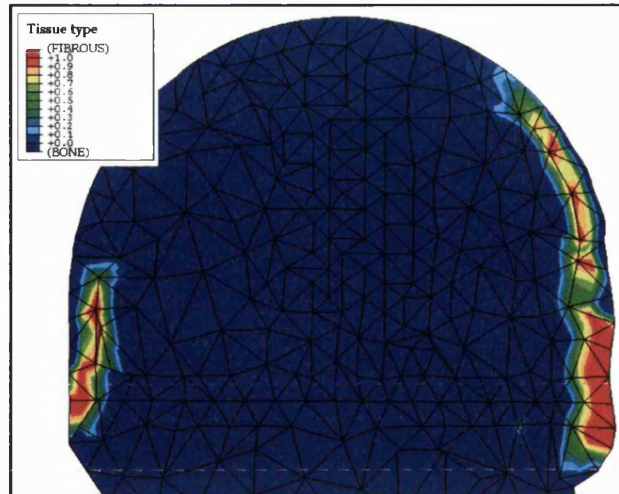


Figure 52: Wagner fibrous tissue simulation (Restricted to interface)

The results of the restricted simulation are shown in Figure 52. The fibrous tissue was seen to advance along the bone interface towards the apex, in a similar manner to that described in clinical failures of the Wagner. However, the fibrous tissue in the simulation was more prevalent on the lateral side, contrary to clinical evidence of more on the medial aspect. It was also worthy of note that some of the material points immediately adjacent to the cement did not turn to fibrous tissue. This suggests that the mesh density was not refined enough to capture the complex behaviour at the interface, and that the local behaviour at the interface may be unduly affecting the wider behaviour of the model, because the elements are so large. It also suggests that the material immediately adjacent to the cement was heavily constrained by the cement. This may be evidence of overconstraint at this boundary or equally it may be evidence of similar behaviour to experimental studies where bone spicules are left intact in the crenellations of the cement due to the strain protection afforded by the microstructure (Mann et al. 1998). However, this behaviour is not appropriate for the current simulation, because of the continuum approach being used, microstructure is not modelled.

Bone/cement interface failure model

Initially, the interface between implant and bone was assumed to be tied. Later in the investigation the effect of possible failure at this interface was investigated. This was carried out because it was known that the cement interdigitation in the Wagner was limited historically. It was also investigated because the behaviour of the fibrous tissue simulation was of interest near the interface.

Interface failure was simulated by including a maximum shear stress parameter in the interface definition. This allowed relative motion between the surfaces if the shear stress at the surface exceeded the parameter, although the surfaces were not allowed to physically separate. It is recognised that this is not realistic interface failure behaviour, but

more realistic modelling, including crack propagation techniques or tensile/shear failure criteria, were considered beyond the scope of the current investigation. The shear stress threshold was set at 2.71MPa using information from studies into the interface behaviour (Mann et al. 1998). If relative motion occurred, the interfaces were assigned a friction coefficient of 0.5. The effect of reducing the shear strength to 0.91MPa to reflect negligible cement interdigitation (Mann et al. 2001) was also investigated.

The results of the simple interface failure model on the fibrous tissue remodelling showed similar behaviour to the patterns seen with the tied interface, except that the rate of fibrous tissue progression was significantly slower (speed approximately halved). Reducing the shear stress threshold further did not seem to have any significant additional effect. When the interface failure model was tested on the bone remodelling analysis, it also had the effect of reducing the rate of remodelling, but with similar patterns of density change. This was thought to be an effect of the slight load redistribution.

Levels of micromotion

The levels of micromotion at the implant interface are of interest, because they are commonly used indicators of implant stability. Therefore it was thought worthwhile comparing the levels of micromotion which occurred at the interface, and the ensuing fibrous tissue which was formed. This could then be compared to experimentally determined levels of micromotion.

The levels of micromotion were examined at the interface of the Alloclassic with and without press-fit (Figure 53). This was done in the absence of fibrous tissue. In the case without press-fit relative micromotion occurred along the majority of the implant edges (up to 25µm) with increased micromotions evident in the proximal half of the stem. The micromotions in the proximo-medial area varied from 25µm to 75µm. The micromotions in the proximo-lateral area varied from 50µm to 140µm.

In the case with press-fit the vast majority of the stem indicates negligible relative motion. A small area of micromotion is evident in the most proximal region, where the levels varied from 10µm to 95µm, suggesting a localised area of deformation. Around 10µm of micromotion was indicated in the most proximal contact region of the lateral side of the implant. A large area of micromotion was also present on the anterior side of the implant (not shown) ranging from 0-8µm at the antero-lateral region in the proximal third of the stem.

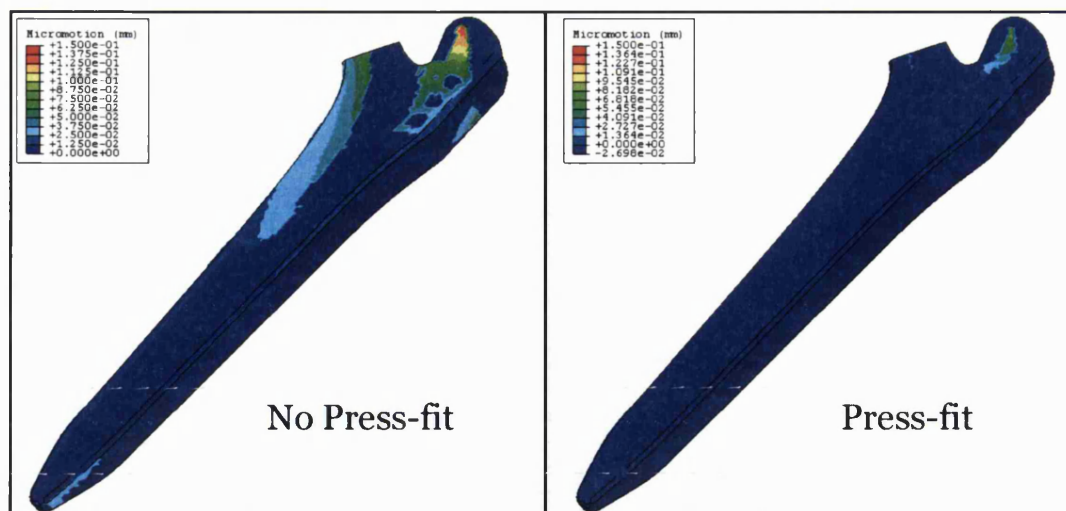


Figure 53: Levels of micromotion the implant interface for the Alloclassic

Götze et al. (2002) measured the primary stability of a series of Alloclassic stems implanted in cadaveric bones under physiological loading. The average relative micromotions from LVDTs varied from 5µm to 34µm. The study by Götze provides some validation of the orders of magnitude of micromotion seen in the FE modelling. The case without press-fit seems to give values significantly higher than those measured by the experimental investigation. Conversely, the case which included press-fit had typical micromotions less than those measured experimentally. It may be the case that for the experimental measurements, the press-fit was lower than that implemented here. It should also be borne in mind that differences may arise from the dissimilar measurement methods used in the FE and experimental techniques. For example, where separation occurs between the implant and bone, no relative motions are available for output. Also, experimental measurements can vary according to the behaviour of external bone reference points, and experimental error.

It is clear from this study that increased press-fit leads to increased implant stability, and decreased relative micromotions. It is also seen that the level of interface micromotion is, in general, proportional to the amount of fibrous tissue produced in these cases. However, the interface micromotion is not necessarily indicative of the level of shear imposed on the surface, so care should be taken in this respect.

Implant fatigue stresses

The levels of stress in the implant are obviously of interest, and also the effect of the simulations on these levels. The von Mises stress is commonly used as an indicator of fatigue performance, in comparison to the theoretical endurance limit of the material under investigation. The fatigue limit for Titanium alloy is typically 610MPa. The von Mises stresses produced under peak Gait loading in the Alloclassic are shown in Figure 54. These indicate a maximum von Mises stress of approximately 350MPa. This occurred in

the lower half of the stem on the lateral side. No significant changes occurred in these implant stresses, after bone remodelling simulation or after the fibrous tissue simulation.



Figure 54: Stress (von Mises) occurring in the Alloclassic stem with press-fit

These results suggest that the majority of the loading is achieved through the distal part of the stem, in contact with the medullary cortical bone. The patterns also indicate some bending stresses are being generated due to the offset in loading. The levels of stress are well below the theoretical endurance limit. This may indicate that there is no risk of fatigue failure, but this does not take account of other environmental factors such as stress raisers or corrosion. There was negligible change in these stresses after the bone remodelling simulation. This indicates that the loss of proximal support, or the effects of proximal fibrous tissue formation are not critical for this design. These findings are in line with the excellent long-term clinical results of this design.

Aspects of pressure induced osteolysis

The occurrence of osteolysis has been associated with the formation of fibrous tissue. It was previously found that mechanical stimulation of the fibrous membrane may be a cause of osteolysis (McEvoy et al. 2002, Rubin et al. 1997), via pressure induced apoptosis of osteocytes in the surrounding bone. For the current simulations it was of interest to see if mechanically induced osteolysis was indicated in any of the cases investigated.

Ideally, to identify the risk of pressure induced osteolysis, the hydrostatic pressure within the tissue should be measured. The pressure could then be compared with levels identified for experimentally induced osteolysis. Identifying this pressure accurately would require biphasic materials descriptions, along with dynamic loading. This is much more complex than the current elastic models, and there is no accurate way of inferring the correct pressures from the elastic models. However, it was thought worth collecting the interface

contact pressures from these elastic models, where fibrous tissue was formed, as this would be a driving factor in the hydrostatic pressure induced in the surrounding tissues. Comparison of contact pressures between different models could then allow ranking of the relative pressures. From this implants could be ranked relative to each other, and the risk be compared with clinical incidences of osteolysis.

Using the current elastic models, contact pressures were measured in areas where fibrous tissue was formed. It was found that the pressures were highly variable locally in each case, and there were no discernible differences in between the pressures seen in the Wagner and Alloclassic cases. Clinically, the Wagner failures are typically attributed to the effects of wear debris related osteolysis. The Alloclassic has also been found to have some incidence of osteolysis in association with proximal fibrous tissue, and also PE wear.

The results of this study did not find any discernible difference in the risks of osteolysis for the different implants. This may be in line with clinical observations of osteolysis in both cases investigated; however, this may also be attributable to the effects of wear debris. The results also indicated that it was difficult to evaluate the risk of osteolysis from the current models, although it may be possible with a much more detailed analysis.

3.8.4 Discussion of results

The investigations into the nature of the shear strain signal indicated that the total octahedral shear strain method of fibrous tissue formation was not a good representation of clinical behaviour. Conversely, the cyclic octahedral shear strain method was shown to be a reasonable representation of clinical behaviour. This indicated that the signal to which the bone responds is likely to be related to the cyclic component of the signal only, such as fluid flow, rather than a damage based signal. This agrees with the findings of Aspenberg et al. (1996), where it was concluded from bone chamber experiments that the formation of fibrous tissue was associated with the parameters of tissue deformation rather than by the amount of tissue damage.

The results of the bone remodelling simulations on the Wagner indicated that the method can produce clinically relevant results. The results of the Wagner simulations also indicated that the orientation/size of the component can affect the results. The bone remodelling results of the Alloclassic with press-fit were very similar to clinical behaviour. The results without press-fit produced results inconsistent with the clinically observed patterns of bone remodelling.

The fibrous tissue simulations on the Wagner were initially similar to clinical behaviour. The progression of the fibrous tissue was then slightly different to clinical behaviour. Further investigations into the behaviour at the interface suggested that the simple model used may not have been refined enough to capture the complex behaviour at the bone/

cement interface, and that the local behaviour at the interface may have been unduly affecting the wider behaviour of the fibrous tissue simulation.

The Alloclassic with press-fit produced a similar pattern of fibrous tissue to those indicated radiologically. The simulation of Alloclassic without press-fit exhibited more fibrous tissue than is typically seen in the clinical situation. For the Alloclassic, increased press-fit was found to lead to increased implant stability, and decreased relative micromotions. It was also seen that the level of interface micromotion was, in general, correspondent to the amount of fibrous tissue produced for the Alloclassic. Loss of proximal support, or the effects of proximal fibrous tissue formation were not found to be critical for the fatigue of the Alloclassic. These findings were in line with the excellent long-term clinical results of the Alloclassic.

The combined bone remodelling and fibrous tissue simulations showed that they can affect each others outcomes. However, the extent of these changes is expected to be design dependent. The combined simulations on the Alloclassic also showed patterns of behaviour similar to that seen in canine in vivo experiments (Jasty et al. 1997), with increases in bone density found immediately adjacent to fibrous tissue. This indicates that the region adjacent to fibrous tissue can receive sufficient mechanical stimulation to initiate bone density increases. It is thought that this behaviour would have a direct effect on the radiographic contrast seen in the tissues, leading to increased detection of radiolucent lines. In a study by Sew Hoy et al. (1981) it was found that animals with a long follow-up with a wide radiolucent line had organisation of woven bone at the interface into horizontal trabeculae. This behaviour in the Alloclassic combined model may be indicative of the of radiolucent lines seen proximally in the Alloclassic radiological studies of clinical follow-up.

Previous applications of bone remodelling and fibrous tissue simulation have been studied separately. Bone remodelling techniques have been used extensively on implant models in the past and have also shown reasonable similarities to clinical behaviour (e.g. Weinans et al. 1993b). Fibrous tissue remodelling, however, is relatively new. The fibrous tissue simulation technique was applied previously by Büchler et al. (2003), in its unmodified form, to an idealised hip implant model. The model provided general agreement with clinical observations of fibrous tissue encapsulation (Büchler et al. 2002, 2003), but specific patterns similar to clinical behaviour were not observed. It was thought that this may have been due to the simplified hip model considered. The application of the model to different qualities of bone density, was in agreement with experimental observations on the effect of bone quality (Büchler et al. 2003). A previous investigation into the failure of resurfacing prostheses by Huiskes et al. (1990) used a simple 2D representation where cement was ignored. To simulate loosening, changes in structure were instigated manually in a similar pattern to clinical observations. The current method has the advantage that it

is not dependant on the user to modify the boundary conditions, but the evolution of the loosening is automatic and dependant on the mechanical environment only.

A common limitation of this, and other remodelling studies of this type, is the simplification of structure and boundary conditions for the purposes of analysis. Discretisation of the complex femoral geometry into elements is routinely carried out. However, due to increases in computer power, the level of refinement has increased with time. The current study is more refined than previous 2D (e.g. Huiskes et al. 1990) and coarse 3D representations (e.g. van Rietbergen et al. 1993). Inherent in most remodelling analyses is the continuum assumption used to represent the detailed structure of bone architecture as a continuum solid. Some recent investigations have managed to overcome this limitation and have modelled small scale, high resolution behaviour of the behaviour of trabecular architecture (Huiskes et al. 2000). However, this level of detail applied over a whole bone model was well beyond the resources of the current investigation.

Remodelling analyses are also typically based on a single static load case (e.g. van Rietbergen et al. 1993), although these methods still demonstrate clinically relevant behaviour. The load case used is generally the peak load occurring during standard walking. This is a simplification of the actual dynamic and continuously varying loading which occurs in vivo. Some authors have overcome this limitation by analysing several different static load cases (Huiskes et al. 1990). Full dynamic analyses are not routinely used due to the additional complexities involved and resource limitations.

The level of press-fit used in the investigation was arbitrary, because no specific information was available regarding the Alloclassic. The actual press fit achieved during surgery may in fact be somewhere between the values investigated here. The press-fit used in the models may be unduly high such that it could cause the bone to exceed yield limits. However, as already mentioned, the post-yield behaviour of bone is currently not very well characterised. The model was considered to be elastic, so behaviour beyond these limits may be unrealistic. However, these results are still thought valuable in relation to the presence (or absence) of press-fit generally, and its overall effects on the patterns of bone remodelling and fibrous tissue formation.

Some of the parameters used in the bone remodelling simulation (such as the lazy-zone) were arbitrary. However, previous bone remodelling investigations on other implants using the same bone remodelling theory have given results which were comparable to clinical evidence (Schmitz et al. 2004, Taylor et al. 2004). In a previous study by Huiskes et al. (1992) it was found that variation of the lazy-zone parameter from $\pm 35\%$ to $\pm 75\%$ could cause magnitudes of bone remodelling to change, but that the overall patterns of bone remodelling were relatively unaffected. Therefore the methods were considered to be appropriate for the current study.

Timescales of fibrous tissue formation were of a similar order to those seen experimentally in canine studies - by definition. However, it was difficult to relate these to timescales for clinical observations in the human case, due to limited information. Threshold and tissue transition rates for canine and human subjects may be significantly different. The timescales of fibrous tissue formation in the case of the Wagner would indicate that loosening failure would progress in a matter of days, and this was not the case clinically. Simulated timescales of fibrous tissue formation should therefore be judged with some caution. However, the patterns of formation are still likely to be informative.

The discrepancies in fibrous tissue patterns from clinical data, for the Wagner, may have been caused by oversimplification of the cement bone interface of the resurfacing implant, or it may be related to the way the fibrous tissue simulation is implemented. The thresholds implemented were assumed as absolute over the physiological levels of signal, in the absence of other information. It may be that the threshold varies spatially, in a similar manner to that implemented in bone remodelling simulations. This hypothesis is supported by the fact that the areas of concern in the model were in areas with physiologically high levels of shear strain, and the areas where the simulation performed well were in areas with physiologically lower values. A possible spatial/donor variation in response to mechanical stimulation was highlighted by the experiments of Sampathkumar et al. (2003). This would support the use of a percentage change in cyclic octahedral shear strain as a more realistic indicator. This would be a worthwhile investigation for future work.

It was also not uncommon for the fibrous tissue simulations to cause the analysis to abort due to excessive deformations in the elements. It should be noted that failure of the analysis is not necessarily an indicator of implant failure, although inferences may be drawn from the excessive deformation. It was recognised that such extreme deformations could lead to anomalous results. These problems are obviously a function of the very low modulus of the fibrous tissue, and the imposed boundary conditions, and are hence unavoidable. This would suggest that the choice of elements for the simulation was not optimal. The element used however, did allow easy integration with current bone remodelling methods, and was hence applicable globally to the whole bone. For further investigations, an alternative and more robust modelling technique should consider these issues. The investigations also indicated that it was difficult to evaluate the risk of osteolysis associated with the fibrous tissue formation, although this may be possible with more complex analyses.

3.8.5 Conclusions

In general, the bone remodelling simulations have provided clinically relevant predictions of density changes which occur in the long-term. The predictions have been accurate for significantly different implants (i.e. uncemented primary stem cf. cemented resurfacing).

These implants have relied on different methods of load transfer and fixation, in different areas of bone.

The fibrous tissue simulations have provided clinically relevant predictions in the short term, for the cases investigated. However, the later progression of the simulation behaviour for the cemented resurfacing prosthesis, was not exactly the same as clinical experience has shown. However, the later progression in the case of the primary uncemented stem was very similar to clinical behaviour.

In the case of the Alloclassic, the results have indicated that press-fit was beneficial to bone remodelling, where more of the bonestock was preserved than without press-fit. The press-fit was also indicated to be beneficial in limiting the amount of fibrous tissue that was produced, by increasing stability. The bone remodelling and fibrous tissue simulation did not appear to impact significantly on the fatigue indications for the stem, and this was borne out by the long term clinical success seen with this implant.

The results of the Alloclassic bone remodelling simulations indicated that the models with press-fit produced more clinically realistic results than those without. This has indicated that press-fit may have an effect on the long-term clinical situation, and should be considered in future simulations. However, the actual level of press-fit which may persist is unknown. The maintenance of press-fit is likely to be reduced over time by creep, damage and bone remodelling mechanisms. Thus, press-fit can be expected to dissipate, but may not disappear completely and may in fact be reinforced by the effect of the taper. It may be the case that other effects may increasingly dominate in the long-term, such as implant integration with bone. The behaviour will no doubt vary on these issues between implant designs.

3.8.6 Testing methodologies discussion

Bone remodelling seems to be an appropriate test for all implants. For stemmed implants this could give an indication of a typical potting level, which would be suitable for fatigue testing. In the case of implants such as the resurfacing implant, it seems this can also be useful in showing how and when failure may occur, and any radiological changes which could be potential indicators of failure, before trauma occurs.

Interface fibrous tissue simulation can also be a useful guide, for predicting the strength and position of initial fibrous tissue formation. Currently the speed and patterns of progression of tissue may not be clinically relevant for cemented interfaces, although some success has been achieved with uncemented interfaces. Further work on this method would obviously be of benefit.

It is realised that whether or not fibrous tissue is predicted is not necessarily an indicator of potential failure. This was highlighted by the Alloclassic and the Wagner simulations.

Fibrous tissue was indicated for both models, and may have been present clinically in both, but the Alloclassic implant was typically highly successful, and the Wagner less so. The same arguments can be stated for bone remodelling predictions. Clearly, these methods are still dependant on the analyst to gauge the risk associated with any behaviour that may be indicated. However, judging the level of risk may be aided by taking account of other relevant factors, such as:

- ▶ the rate of fibrous tissue progression (secondary stability)
- ▶ the amount of initially indicated fibrous tissue (primary stability)
- ▶ the position of fibrous tissue (proximity to wear debris generating site)
- ▶ the proportion of interface fibrous/remaining bone (stability)
- ▶ the increase in implant stress due to bone remodelling (worst case loading)

Many factors could be analysed for bone remodelling and fibrous tissue simulations, both individually and combined together. These factors could then be compared with analyses of other implants, which can draw on significant clinical experience. From such comparative studies, for example, it could be gauged whether an implant is more or less likely to suffer from secondary stability issues, than one which is currently in service.

3.8.7 Summary

The aim of this investigation was to implement the bone remodelling and fibrous tissue simulations on FE models of the Wagner and Alloclassic implants, and then compare the results with clinical behaviour of these implants. This was then used to judge whether these methods are suitable for use as part of a pre-clinical testing strategy, and to help evaluate their roles in this field.

Bone remodelling and fibrous tissue simulations were implemented on the cemented Wagner resurfacing prosthesis and the uncemented Alloclassic primary stem. The bone remodelling simulations provided excellent agreement with clinical behaviour. The fibrous tissue simulations provided initial agreement with clinical behaviour, although there was some deviation from clinical behaviour in later patterns near the cemented interface.

These simulation techniques were thought useful as part of a pre-clinical testing strategy, although the levels of risk associated with simulation results still requires interpretation by the analyst. However, judging the level of risk can be greatly enhanced with the judicious use of these tools.

3.9 Pre-clinical test methods for new implants

3.9.1 Aim

The aim of this investigation was to implement the pre-clinical testing strategies on newer implants with a view to identifying possible failure scenarios.

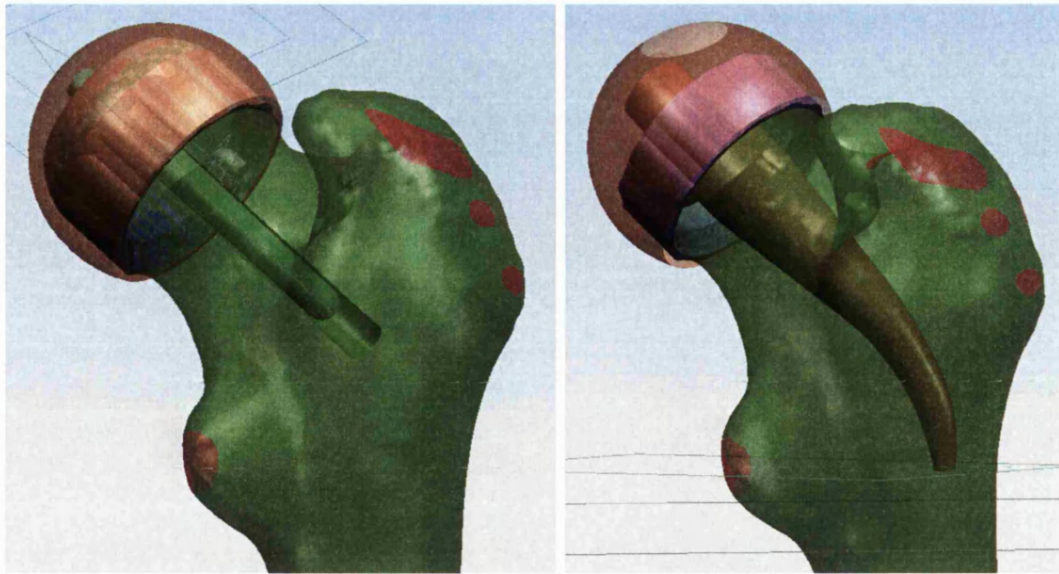
3.9.2 Investigation outline

The chosen implants for this investigation were the Durom resurfacing implant, and the Durom-Hybrid (Zimmer GmbH). The Durom is a stemmed type of resurfacing implant, in current and increasing usage, but for which no long-term clinical information was currently available. However, in early results of 200 Durom hips (average 26 months) there was no evidence of failure, component migration or osteolysis (Grigoris et al. 2005). The Durom-Hybrid is a novel, brand-new implant which is a cross between a resurfacing design and a short stemmed transtrochanteric design.

Bone remodelling and fibrous tissue simulations were carried out on models of the Durom and Durom-Hybrid. Clinical evidence from similar implants was then evaluated and compared to the predicted behaviour. Possible failure scenarios were evaluated and further strategies suggested where appropriate.

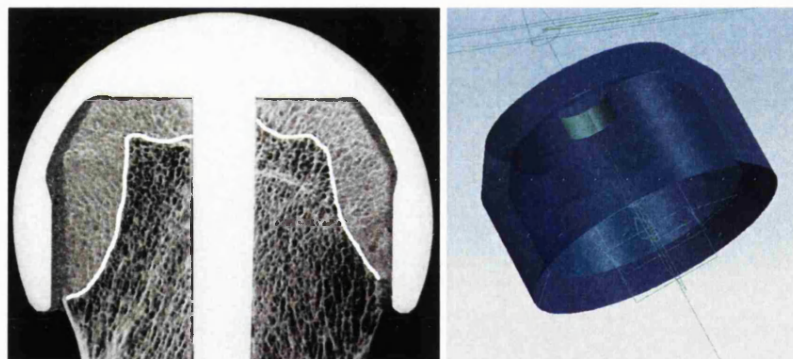
3.9.3 Modelling

The bone geometry was the same as that used in the previous investigations. Meshing methods and materials properties were as described in previous investigations. The implant models, and custom virtual implantation tooling solids were created in Unigraphics. The Durom (Figure 55a) and Durom-Hybrid (Figure 55b) were both implanted coaxial with the neck, following typical surgical procedure. All small features (<1mm) were removed which could unduly affect the meshing process. The Durom was modelled as Cobalt Chrome head and stem. The Durom-Hybrid was modelled as a Cobalt Chrome head, with a Titanium stem. The Durom-Hybrid was modelled with press-fit along the tapered part of the stem, at a similar level to that used previously.



Figures 55a & 55b: Durom and Durom-Hybrid models respectively

Cement was modelled at a depth of 0.75mm in both models. Most surgeons now use jet lavage, which leads to large cement penetration during surgery. To take account of this behaviour cement penetration was included in the models. An experimental study into typical cement penetrations, by Howald et al. (2005), was used as a guide to create cement penetration geometry for this study, as shown in Figures 56a & 56b.



Figures 56a & 56b: Cement penetration (experimental and modelled respectively)

Investigation into the behaviour of cement penetrations by Jofe et al. (1991) found that penetration depth increased with applied pressure, time of pressure application, lower cement viscosity and greater porosity of bone. The tensile and shear strength of the composite have also been shown to increase with the depth of cement penetration, better cleaning and earlier insertion of cement. The area of penetration is essentially a three phase composite of bone, cement and air porosities. Studies with bovine bone indicated that the standard 'method of mixtures' could not be used to predict the properties of the composite.

Jofe et al. (1991) mechanically tested human tibia bone composites and found that the compressive modulus using hand filled LVC at 2 minutes was 1210 MPa. It was also found that there was no correlation of properties with the density of the bone – contrary to previous studies with bovine bone. Williams and Johnson (1989) investigated the mechanical properties of bovine tibial trabecular bone composites and found that the dynamic Poisson's ratio did not seem to vary with bone volume fraction. Therefore, for the current investigation, the cement/bone composite was assumed to have mechanical properties of $E=1210\text{MPa}$, $\nu=0.3$. The interfaces between cement/composite/bone were considered to be tied. Other interface characteristics (cement/implant, bone/implant) were identical to those used previously.

Muscle and femoral head loading was identical to that used in previous investigations. The implant load application for the resurfacing prostheses was based on the subtended angle of contact derived from values found by Udofia and Jin (2003), who analytically modelled the contact mechanics of typical metal-on-metal resurfacing prostheses. Readings were taken of the subtended angle of the pressure profile over the boundary lubrication film. From this information a similar subtended angle was used for the metal-on-metal implants under investigation, and an appropriate circular contact radius was derived.

3.9.4 Simulation results

The bone remodelling simulation results for the Durom indicated a significant area of bone loss under the superior region of the cup which was typically between 40% and 60% bone loss for the 10 year simulation (Figure 57). Small areas of positive and negative bone density changes were also seen around the stem. These changes occurred due to the redistribution of stresses around the stem hole, rather than through stem loading itself. No stem/bone contact occurred during the simulation, consistent with the design intent of the implant.

The fibrous tissue simulation results for the Durom are shown in Figure 58. The fibrous tissue developed in a similar manner to the Wagner simulations. The fibrous tissue initially formed around the periphery of the cup. The progression was initially along the interface between the bone and cement/bone composite, and then moved across the neck. The simulation failed due to excessive deformations after 12 steps (simulated 12 days).

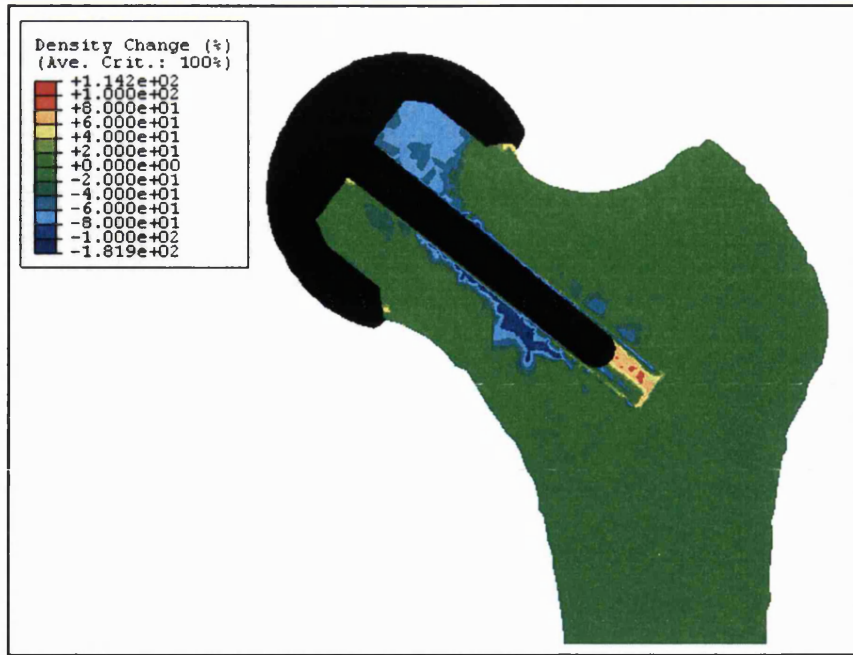


Figure 57: Percentage change in bone density for Durom bone remodelling

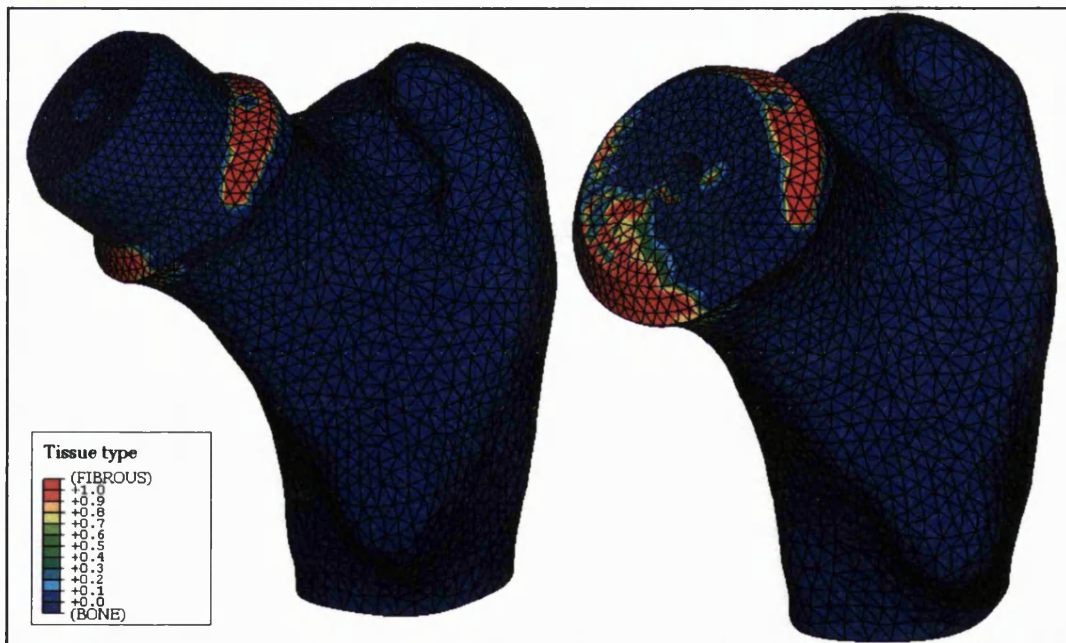


Figure 58: Fibrous tissue produced for Durom resurfacing

The bone remodelling simulation for the Durom-Hybrid (Figure 59) indicated that the majority of the stem would stimulate increased bone density in the area surrounding the stem (typically around 40% to 60% bone density increase). There was also some indication of significant bone loss under the cup, especially under the medial side of the implant which had lost of over 80% bone density.

The fibrous tissue simulation for the Durom-Hybrid indicated that fibrous tissue would form at the lateral tip of the stem, near the bottom quarter of the stem (Figure 6o). An additional small amount of fibrous tissue was indicated more proximally on the medial side of the stem. No fibrous tissue was indicated near the edge of the articulating surface.

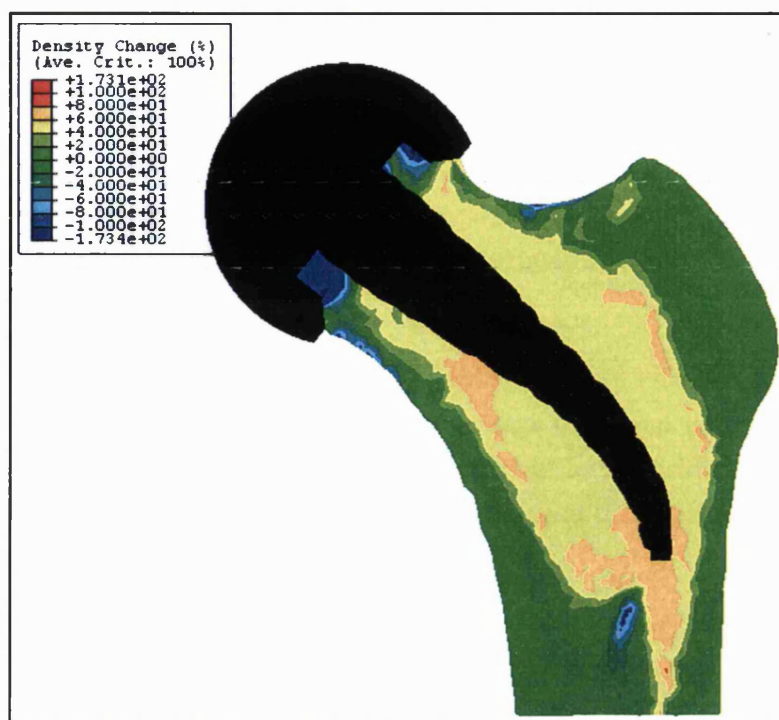


Figure 59: Percentage change in bone density for Durom-Hybrid bone remodelling

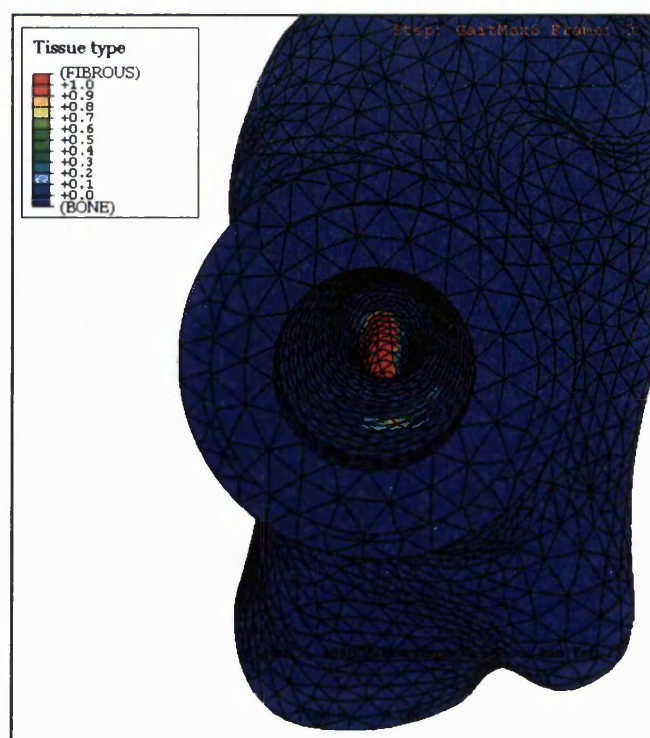


Figure 60: Fibrous tissue produced for Durom-Hybrid

3.9.5 Clinically similar results

Designs similar to the Durom

Nelson et al. (1997) detailed the histology of a very short stemmed, cemented hemi-resurfacing arthroplasty. Resected femoral heads showed a membrane between the bone/cement interface. The cement was intact, and although the peripheral edges of the remnant of the femoral head contained necrotic segments. There was no impregnation of metallic debris into the head itself.

In a study of the Conserve Plus (Amstutz et al. 2004), a cemented resurfacing implant with a short metaphyseal stem to facilitate accurate component alignment, at clinical follow-up substantial radiolucencies were seen around 16/400 of the uncemented metaphyseal stems.

In a follow-up study by Campbell et al. (2004) of the Conserve Plus cemented resurfacing component, with cemented/press-fit stem, the average follow-up was 35 months in the series of 400. Femoral loosening was the main cause of failure with 7 revisions. Loss of femoral head bone was found to occur with fibrous tissue at the interfaces, consistent with bone cement debris.

Kishida et al. (2004) carried out a radiological study of the BHR. DEXA measurements were taken near the stem after 24 months implantation. Where changes did occur, there were increases in density around the stem.

In another study of the BHR (McMinn et al. 1996) at 3 years 26% of femoral components had a surrounding radiolucent line, and mean femoral neck thinning was 1.47mm. Interface fibrosis was noted in loosened implants.

Designs similar to the Durom-Hybrid

The TARA is an uncemented prosthesis with a long thin angulated stem. Mallory et al. (1984) detailed an average follow-up of 2 years with the TARA in 60 patients. There was progressive occurrence of bone sclerosis (55% at 2 years cf. 75% at 3 years) along the lateral border of the prosthetic stem, usually at the distal tip, indicating that the stem provoked bone changes in response to stress transfer. If the stem contacted the cortex, sclerosis was consistently present.

Head (1984) examined the clinical behaviour of 64 TARA prostheses, and there was some evidence of proximal stress shielding caused by the load transfer through the stem. It was said that initial results were better than the Wagner but long-term results were worse.

De WaalMalefijt and Huiskes (1993) carried out a radiological investigation of 72 TARA prostheses. In some of the implants a sclerotic line was seen around the femoral stem. This line was observed on the medial side in 36% of the hips, on the lateral side in 55%

and around the tip in 67%. In some cases there was also evidence of radiolucency between this sclerotic line and the stem.

3.9.6 Discussion and recommended testing strategies

The level of press-fit used in this investigation was arbitrary for the tapered section of the implant. The actual press fit achieved during surgery may be different and will depend on the surgical procedure. Some of the simulation parameters were arbitrary although have given reasonable results in previous investigations. As in found in the previous investigations there was limited confidence in the accuracy of time based parameters, although the simulations were thought reasonable in terms of general patterns of behaviour. By the nature of the implants under investigation there was limited availability of clinical data for comparison purposes, and additional similar implants have been investigated. This itself imposes a limitation of confidence in the clinical relevance of the predictions.

The simulation results for the Durom seem to indicate that fibrous tissue formation can occur in a similar manner to the Wagner (i.e. around the periphery of the implant). This seems reasonable, as the implants are substantially the same, and the stem is not intended to be load bearing. It may be the case, that the PE wear debris substantially accelerated this failure in the case of the Wagner. This may not necessarily occur with the Durom, as it is a metal-on-metal bearing. The bone remodelling simulations, also have similarities to the Wagner, although the stem does alter the load pattern slightly. Bone loss is seen to be mainly restricted to the lateral side of the head, whereas the medial side receives additional loading from the proximal part of the stem, and does not lose significant density. Clinical results for similar implants seem to substantially agree with the predictions of fibrous tissue, and some of the bone remodelling changes seen around the stem. However, the bone remodelling changes are expected to be dependant on whether the stem is load bearing (intentionally or otherwise). Early clinical results of the Durom itself show no signs of failure (Grigoris et al 2005).

Recommended strategies for the Durom are limited, as radiological follow-up is impeded by the radio-opacity of the implant cup which obscures the region of interest under the cup. Monitoring of the implant stem for significant changes in density could also be useful, as this could indicate non-design loading. It is thought that further fatigue testing procedures would not be appropriate for this implant, as the predicted changes would lead to symptomatic implant loosening requiring revision. However, the revision and conversion to a standard primary procedure would be relatively simple due to the limited volume of bone affected.

The simulation results for the Durom-Hybrid indicate some proximal bone loss, and significant distal bone density increases. This may indicate that stress-bypass could be an

issue for this implant if proximal bone losses spread significantly. The simulations also predicted fibrous tissue formation in the distal part of the stem. Clinical evidence of proximal stress shielding and distal sclerosis in other similar designs showed some similarities to the predicted behaviour. It should be noted that the fibrous tissue indicated was remote from the main wear debris generating site, so the chances of this membrane becoming osteolytic may be reduced compared to standard resurfacing implants.

A recommended strategy for the Durom-Hybrid would be cantilever fatigue testing, as this implant may be affected by a stress-bypass phenomenon similar to traditional stems. The potting level for this fatigue test may be judged from the indications of the bone remodelling study. An additional predictive study, without press-fit, should also be consulted, in the event that press-fit is not maintained. It would also be advisable to check for proximal bone loss and distal sclerosis, which could be evidence of stress bypass, during clinical trials. Checking for evidence of distal radiolucencies would also be advisable, as these may precede loosening of the implant.

3.9.7 Investigation Summary

The aim of this investigation was to implement the numerical pre-clinical testing strategies on relatively new implants with a view to identifying possible failure scenarios. Bone remodelling and fibrous tissue simulations were carried out and the simulation behaviour was compared with the clinical behaviour of similar implants. Possible failure scenarios were identified, and pre-clinical testing and monitoring strategies were suggested based on the results.

3.10 Summary of predictive behaviour investigations

The aim of this investigation was to focus on how implants interact with the biological host bone and cause it to evolve with time. The objective was to describe the failure processes using numerical techniques, and ultimately predict potential failure mechanisms, as a tool to aid pre-clinical testing and to enable greater understanding of the failure processes.

The sequence of events which typically cause loosening was investigated. Bone remodelling and fibrous tissue formation were explored, and suitable simulation techniques were developed. These methods were then evaluated on a resurfacing device and a primary stem with significant clinical experience. The relevant clinical evidence for the chosen implants was evaluated. Comparison of simulation behaviour with clinical evidence yielded reasonable results in the majority of cases. The investigation identified the importance of press-fit for both bone remodelling and fibrous tissue formation. The simulation methods were thought useful as a pre-clinical testing tool in themselves. These methods were then tested on relatively new implants, with a view to providing useful information for further pre-clinical testing strategies and radiological follow-up techniques.

4 Conclusions and recommendations for further work

4.1 Summary

The aim of this investigation was to develop suitable pre-clinical testing strategies for hip implants and elucidate risk factors that may lead to accelerated prosthetic failure.

Initially, standard pre-clinical testing methods were evaluated. It was found that testing strategies typically require prior knowledge of any worst case loading scenario. This worst case scenario may vary between designs. It was thought that the current pre-clinical testing standards were inadequate for the evaluation of novel designs, due to the inherent lack of clinical experience.

A comparative design investigation was used to explore the behaviour of the different groups of conservative implants, as these were not covered by existing pre-clinical testing standards. A broad range of conservative implants were classified by their overall geometric design. The different groups were then evaluated in a comparable manner using FEA. The results showed patterns of behaviour that were comparable to the available clinical evidence. The investigation showed that different lengths of stem may be prone to different types of failure. The more sinister type of failure scenarios seemed to be associated with the threat of loosening and bone remodelling.

Subsequent work investigated the prediction of potential failure scenarios which may occur in the future, to aid in pre-clinical testing. Bone remodelling and fibrous tissue formation were explored, and suitable simulation techniques were investigated. These methods were then evaluated using a resurfacing device and a primary stem with significant clinical experience. Comparison of simulation behaviour with clinical evidence yielded reasonable results in the majority of cases. The investigation was also useful in identifying the role of variables such as press-fit. The simulation methods were thought useful as a pre-clinical testing tool in themselves. These methods were then tested on relatively new implants, with a view to providing useful information for further pre-clinical testing strategies and radiological follow-up techniques.

4.2 Evaluation of objectives

There now follows a brief description of the investigation performance in relation to the previously stated objectives:

To investigate the effects of implant design, using FE and clinical evidence

The finite element method was applied to various classifications of implant design and clinical evidence was also evaluated for these groups. The immediate postoperative state was modelled using FE and the risk of fatigue failure was found to be similarly low for all designs. The FE simulations indicated that longer implants were more at risk of the stress bypass phenomenon, and this was also seen in the clinical evidence. In the FE implant length was not found to correlate directly with primary stability, although the effect of a collar was significant. In clinical evidence, the biological changes seen subsequent to implantation (fibrous tissue formation and bone remodelling) were thought to be of greater concern for implant longevity than the case in the immediate postoperative state.

To investigate the biomechanical interactions which could result in implant failure through evolution of the biological structure

The available evidence regarding implant loosening behaviour was rationalised into a reasoned explanation of the mechanical processes behind implant loosening through fibrous tissue formation. Comparison of subsequent FE investigations with clinical behaviour revealed insight into the nature of the signalling mechanism involved in fibrous tissue formation. It was found that mechanical initiation of fibrous tissue formation was more likely to be related to fluid flow mechanisms than damage based mechanisms.

To develop and combine numerical methods to simulate possible failure behaviour

Bone remodelling and fibrous tissue formation algorithms were developed based on current understanding and combined with the finite element method, to simulate future behaviour of the biomechanical system. The simulations were implemented in such a way that both simulations could be investigated on the same model. Subsequently it was found that the choice of elements may not have been ideal for the resultant high strain situations, but the method did allow coherent combination of the two simulation techniques to the whole of the bone model.

To validate numerical simulation methods using clinical evidence

The numerical simulation results have been compared to relevant clinical data for different implants. The bone remodelling simulations have shown patterns of behaviour similar to clinical behaviour for both a cemented resurfacing and an uncemented primary stem. The simulation of fibrous tissue was clinically relevant for the uncemented stem, but was less so for the cemented resurfacing implant. Nevertheless, there were still reasonable similarities to clinical behaviour, but the method may benefit from further refinement.

The combination of methods was seen to have some effects on results of the simulations, but did not interact to change overall results significantly in the case investigated. The results of the combined simulations did however, indicate additional clinically relevant behaviour, with increases in bone density predicted immediately adjacent to fibrous tissue which had formed, similar to behaviour which has been observed in other in vivo experiments.

To model the consequences of implantation parameters on predictive simulations

Press-fit was found to be an important factor, because it had a significant effect on the performance of the bone remodelling and fibrous tissue simulations. Press-fit was found to increase implant stability, and also reduce the extent and rate of progression of fibrous tissue formation during simulations. Press-fit was also found to increase the loading in the surrounding bone leading to reduced bone loss and even increases in bone density during bone remodelling simulations.

To determine the effect of implantation parameters in the clinical situation

Comparison with clinical behaviour of simulation results with press-fit yielded more realistic representations than the results without press-fit. This was seen as evidence that if press-fit is used during the surgical procedure, it may in fact persist long-term in the clinical situation. However, the persistence of press-fit was also expected to be design dependant, with factors such as bone remodelling and levels of interface disruption being important.

To gain insight into the effects of biological phenomena on the failure of femoral implants

The occurrence and progression of bone remodelling and fibrous tissue has been observed for simulations on different implants. A greater understanding of the mechanical reasons and consequences of observed clinical and radiological behaviour has been developed. The extent of fibrous tissue formation was found to be related to the stability of the implant. From the simulation results, the appearance of significant radiolucent lines in the long-term has been related to fibrous tissue formation and concomitant bone remodelling due to increased strain in the surrounding bone. However, this was not necessarily an indicator of clinical failure. The effects of predicted biological evolution on the risk of failure of an implant can vary between implant designs, and needs to be interpreted carefully.

To develop methods as an aid to pre-clinical testing

The numerical methods developed here are useful tools for the simulation of potential future biological evolutionary behaviour for any design of implant. The information from simulations has been used in combination with available clinical data and simulations for

other implants to gauge the risk associated with novel designs, and suitable mitigation recommended. These methods are a useful aid to pre-clinical testing strategies where limited relevant clinical experience is available.

4.3 Conclusions

A pre-clinical implant design investigation technique has been developed based on current understanding of biomechanical processes and implemented using numerical methods to simulate potential future behaviour of implants. The methods have shown similar behaviour to clinical evidence, which is promising, although further validation of these methods would obviously be beneficial. The methods can indicate possible effects of implants on both bone remodelling and fibrous tissue formation. This behaviour can then be compared with relevant clinical evidence, and simulations of other implants to gauge the risk associated with potential failure scenarios. The simulation results can then also be used for designing further testing or follow-up methods to mitigate this potential behaviour, or even aiding in redesign of the implant in question.

4.4 Recommendations for further work

From the current investigation, there are several areas which have been identified for further investigation, which could be of additional scientific benefit.

During the FE investigations two different signal types were investigated which produced similar magnitudes of fibrous tissue. However, under the effects of press-fit, the simulations produced opposing patterns of behaviour. These results present the opportunity to understand the nature of the signal in more depth. Experimentation on the level of press-fit under displacement control, may help distinguish more decisively whether the fibrous tissue initiating signal is damage or fluid flow based. The experiment could be similar in nature to previous canine studies investigated, with an additional variation in press-fit prescribed. These experiments could then be modelled and simulated in FE for an enhanced understanding of the processes occurring.

The fibrous tissue simulation used the thresholds for tissue formation at absolute levels, over the levels seen physiologically. Another way identified for implementing the thresholds would be to use a percentage change from the physiological level. This would also take account of the variations seen locally in physiological levels of shear, in a similar manner to current empirical bone remodelling theories. This may provide more realistic simulations of fibrous tissue, especially in the femoral neck area where high shear strains are typically seen. This approach may also be indicated by a possible spatial/donor variation in response to mechanical stimulation which was highlighted by the experiments of Sampathkumar et al. (2003). The effect of variation of this threshold may also be of interest, as it can be expected to vary between species.

During the investigation, the areas typically comprised of trabecular arcades in the bone were found to be associated with high levels of shear, and also the external bone ridges were seen to be associated with regions of high surface stress. Further investigation into the form/function relationships of bone may be of interest in terms of the regional variation of the structural optimisation. The ability of different forms of bone could be tested experimentally to establish the effect of form on strength under different types of loading.

It was thought that further investigations into fibrous tissue simulation should use more robust modelling of soft tissues, as the large distortions have been found to be problematic. It would also be useful to carry out more detailed modelling. A more detailed interface failure model may also enable behaviour at the cement/bone interface to be modelled more realistically. A biphasic model would also enable investigation into possible osteolytic pressure generation from fibrous tissue.

Lastly, a clinical study by Wick and Lester (2004) concluded that small variations in external configuration for two different designs could have a statistically significant effect on radiological consequences. Leading on from this, it would be worthwhile investigating if the current simulation techniques could distinguish between the predicted behaviours for the two different designs cited in this clinical study. This could enable a greater understanding for the reasons for the differences seen, and would also help establish further the level of veracity of the simulation methods.

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Appendix A: FORTRAN subroutines

For the numerical simulations, two different FE models were used for each implant design investigated. The natural, unimplanted model was required as a reference against which changes from the physiological normal behaviour could be measured. The implanted case is the model in which the bone remodelling and fibrous tissue formation were simulated.

Reference model algorithm

The reference model algorithm is made up of two subroutines which interface with the ABAQUS simulation of the natural unimplanted femur under physiological loading. The first subroutine (UEXTERNALDB) writes the required information to data files at the end of the simulation. The second subroutine (UARM) is used to calculate user defined output variables from the analysis (e.g. Strain Energy Density), and material orientation information. This information is passed to the UEXTERNALDB subroutine for writing. The material cards for the reference model are all required to call 2 user output variables to utilise this routine.

Parameters which need to be adapted to suit each individual analysis are NUMEL (number of elements in the model) and NUMOID (number of material orientation definitions). Arbitrary large numbers may be used, although this may impact significantly on analysis performance. The numerical suffixes which are typically appended to consecutive material and orientation names (e.g. MAT01, MAT02, MAT03 ...) are used to advantage in this algorithm, using character substrings to extract the numerical suffixes. Thus the exact form of the substring will depend on part and material names, and should be altered accordingly.

Start of reference model algorithm

```
-----
C      This subroutine writes out the information from the reference model
C      (i.e. state before implantation)
C
CCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCC
SUBROUTINE UEXTERNALDB(LOP,LRESTART,TIME,DTIME,KSTEP,KINC)
  INCLUDE 'ABA_PARAM.INC'
  DIMENSION TIME(2)
  CHARACTER*256 OUTDIR,JOBNAME,REFFILE,OIDFILE
  INTEGER NUMEL,NUMOID
  PARAMETER (NUMEL=77000,NUMOID=298)
  INTEGER MATOID(NUMEL)
  DIMENSION DAT(NUMEL,6,4), TEE(NUMOID,3,3)
  COMMON/KBLOCK/DAT,TEE,MATOID
  SAVE /KBLOCK/

C      At end of analysis write out arrays to data files
  IF (LOP.EQ.3) THEN
    CALL GETOUTDIR(OUTDIR,LENOUTDIR)
```



```

        DAT(NOEL,3,NPT) = COORD(3)
ENDIF

C      Convert the material name suffix (character variable) to a
C      justified integer variable using appropriate substrings
      WRITE (TEMPA,'(A2)') CMNAME(4:5)
      READ (TEMPA,'(BNI2)') MATNO

C      Calculate material density from defined method
      IF (MATNO.EQ.1) THEN
        DAT(NOEL,5,NPT) = 174.D0
      ELSE
        DAT(NOEL,5,NPT) = (MATNO*50.D0-75.D0)*0.872D0 + 174.D0
      ENDIF

C      Calculate Strain Energy Density from model (SED)
      CALL GETVRM('ENER',ARRAY,JARRAY,FLGRAY,JRCD,JMAC,JMATYP,MATLAYO,
1 LACCFLA)
      UVAR(1) = ARRAY(1)/DAT(NOEL,5,NPT)
      DAT(NOEL,4,NPT) = UVAR(1)

C      Calculate Octahedral Shear Strain (Buchler definition)
      CALL GETVRM('EEP',ARRAY,JARRAY,FLGRAY,JRCD,JMAC,JMATYP,MATLAYO,
1 LACCFLA)
      UVAR(2) = (1.0D0/DSQRT(3.0D0))*DSQRT( (ARRAY(1)-ARRAY(2))**2.0D0
1      +(ARRAY(2)-ARRAY(3))**2.0D0 +(ARRAY(3)-ARRAY(1))**2.0D0)
      DAT(NOEL,6,NPT) = UVAR(2)

C      Convert the material orientation suffix (character variable) to a
C      justified integer variable using appropriate substrings
      WRITE (TEMPB,'(A3)') ORNAME(27:29)
      READ (TEMPB,'(BNI3)') MATOID(NOEL)

C      Define orientation array
      TEE(MATOID(NOEL),1,1) = T(1,1)
      TEE(MATOID(NOEL),2,1) = T(2,1)
      TEE(MATOID(NOEL),3,1) = T(3,1)
      TEE(MATOID(NOEL),1,2) = T(1,2)
      TEE(MATOID(NOEL),2,2) = T(2,2)
      TEE(MATOID(NOEL),3,2) = T(3,2)
      TEE(MATOID(NOEL),1,3) = T(1,3)
      TEE(MATOID(NOEL),2,3) = T(2,3)
      TEE(MATOID(NOEL),3,3) = T(3,3)

C      Debug to the .msg file
      IF ((NOEL.EQ.7).AND.(NPT.EQ.1)) THEN
        WRITE (7,*) 'DEBUG OUTPUT FOR ELEMENT 7 NPT 1'
        WRITE (7,*) 'UVARM WAS CALLED AT KSTEP',KSTEP,' KINC',KINC
        WRITE (7,*) 'DAT(NOEL,1,NPT) IS',DAT(NOEL,1,NPT)
        WRITE (7,*) 'DAT(NOEL,2,NPT) IS',DAT(NOEL,2,NPT)
        WRITE (7,*) 'DAT(NOEL,3,NPT) IS',DAT(NOEL,3,NPT)

```

```

WRITE (7,*) 'UVAR(1) IS',UVAR(1)
WRITE (7,*) 'UVAR(2) IS',UVAR(2)
ENDIF

RETURN
END

```

End of reference model algorithm

Remodelling algorithm

The remodelling algorithm is made up of four subroutines which interface with the ABAQUS simulation of the implanted femur under physiological loading. At the beginning of the analysis, the first subroutine (UEXTERNALDB) reads in the information from data files created from the reference case. The second subroutine (ORIENT) uses this information to set the initial material orientations for the material points. The assignment of material orientations in this way is specified via user defined material orientations in the material card. This method ensured that identical initial conditions were created to the reference case. The method of matching between models was based on minimisation of the distance between material points, before the model is loaded. This method was chosen because internal element labels can vary between models, and global element number retrieval was non-trivial.

The third subroutine (SDVINI) sets the initial values of the solution dependant variables. These variables are used to control the evolution of material properties throughout the simulations, and they are set the materials properties at the start of the analysis. The values of the solution dependant variables are based on data from reference case, and implemented in a similar manner to the ORIENT subroutine.

The fourth subroutine (USDFLD) contains the main remodelling code. This subroutine monitors the values of the solution dependant variables throughout multiple steps of the remodelling analysis, and alters the values of the two field variables accordingly. These field variables are used to control the mechanical properties of the bone throughout the analysis, simulating changes in structure over time. The subroutine interfaces with the model via the use of two field variables and eight solution dependant variables. The first field variable controls the bone density, and the second field variable controls the formation of fibrous tissue. The material card for the bone hence takes the following form:

```

*Material, name=Bone
*Depvar
8,
*Elastic, dependencies=2, type=ENGINEERING CONSTANTS
0.1, 0.1, 0.1, 0.42, 0.23, 0.23, 0.1, 0.1
0.1, , 0., 0.

```



```
C      Debug to .msg file
      IF ((NOEL.EQ.9969).AND.(NPT.EQ.1)) THEN
          WRITE (7,*) 'ORIENT HAS BEEN CALLED for NOEL 9969 NPT 1'
      ENDIF

      RETURN
      END
```

```

SUBROUTINE SDVINI(STATEV,COORDS,NSTATV,NCRDS,NOEL,NPT,
1  LAYER,KSPT)
  INCLUDE 'ABA_PARAM.INC'
  DIMENSION STATEV(NSTATV),COORDS(NCRDS)
  INTEGER NUMINT,N,KFLAG
  PARAMETER (NUMINT=308000)
  DIMENSION DAT(NUMINT,6)
  COMMON/KBLOCK1/DAT
  SAVE /KBLOCK1/

```

```
C      Assign initial solution dependant data for material point
STATEV(1) = DAT(KFLAG,4)
STATEV(2) = DAT(KFLAG,5)
STATEV(3) = 0.D0
STATEV(4) = STATEV(2)
STATEV(5) = 0.D0
STATEV(6) = DAT(KFLAG,6)
STATEV(7) = 0.D0
STATEV(8) = 0.D0
```



```

C      Debug to the .msg file
        IF ((NOEL.EQ.9969).AND.(NPT.EQ.1)) THEN
            WRITE (7,*) 'FIBROUS AT KSTEP',KSTEP,'KINC',KINC
        END IF

C      Define nature of driving signal
        SIGNAL= STATEV(8)-STATEV(7)-STATEV(6)

C      Define Driver function - M(epsilon)
        IF (SIGNAL.LT.EMIN) THEN
            DRIVER= 0.D0
        ELSE IF (SIGNAL.GT.EMAX) THEN
            DRIVER= 1.D0
        ELSE
            DRIVER= (SIGNAL-EMIN)/(EMAX-EMIN)
        END IF

C      Define DeltaKappa
        DELTA = -WEEKS*TNU*(STATEV(5)-DRIVER)

C      Remodel Kappa
        STATEV(5) = STATEV(5) + DELTA

C      End of interface remodelling
        END IF

C      End of dual remodelling
        END IF

C      End of global controls
        END IF

C      Redefine Field variables
        FIELD(1) = (STATEV(4)**2.D0)/(RHO**2.D0)
        FIELD(2) = STATEV(5)

C      Debug to the .msg file
        IF ((NOEL.EQ.9969).AND.(NPT.EQ.1)) THEN
            WRITE (7,*) 'USDFLD HAS BEEN CALLED AT KSTEP',KSTEP,'KINC',KINC
            WRITE (7,*) 'FOR NOEL 9969 NPT 1'
            WRITE (7,*) 'STATEV(1) IS',STATEV(1)
            WRITE (7,*) 'STATEV(2) IS',STATEV(2)
            WRITE (7,*) 'STATEV(3) IS',STATEV(3)
            WRITE (7,*) 'STATEV(4) IS',STATEV(4)
            WRITE (7,*) 'STATEV(5) IS',STATEV(5)
            WRITE (7,*) 'STATEV(6) IS',STATEV(6)
            WRITE (7,*) 'STATEV(7) IS',STATEV(7)
            WRITE (7,*) 'STATEV(8) IS',STATEV(8)
            WRITE (7,*) 'FIELD(1) is',FIELD(1)
            WRITE (7,*) 'FIELD(2) is',FIELD(2)
        END IF

```

END IF

RETURN

END

End of remodelling algorithm